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# (54) ANALYTE SENSOR

ANALYTSENSOR

CAPTEUR D'ANALYTE

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## Description

### FIELD OF THE INVENTION

**[0001]** The preferred embodiments relate generally to systems and methods for measuring an analyte in a host.

### BACKGROUND OF THE INVENTION

**[0002]** Diabetes mellitus is a disorder in which the pancreas cannot create sufficient insulin (Type I or insulin dependent) and/or in which insulin is not effective (Type 2 or non-insulin dependent). In the diabetic state, the victim suffers from high blood sugar, which can cause an array of physiological derangements associated with the deterioration of small blood vessels, for example, kidney failure, skin ulcers, or bleeding into the vitreous of the eye. A hypoglycemic reaction (low blood sugar) can be induced by an inadvertent overdose of insulin, or after a normal dose of insulin or glucose-lowering agent accompanied by extraordinary exercise or insufficient food intake.

**[0003]** Conventionally, a person admitted to a hospital for certain conditions (with or without diabetes) is tested for blood sugar level by a single point blood glucose meter, which typically requires uncomfortable finger pricking methods or blood draws and can produce a burden on the hospital staff during a patient's hospital stay. Due to the lack of convenience, blood sugar glucose levels are generally measured as little as once per day or up to once per hour. Unfortunately, such time intervals are so far spread apart that hyperglycemic or hypoglycemic conditions unknowingly occur, incurring dangerous side effects. It is not only unlikely that a single point value will not catch some hyperglycemic or hypoglycemic conditions, it is also likely that the trend (direction) of the blood glucose value is unknown based on conventional methods. This inhibits the ability to make educated insulin therapy decisions.

**[0004]** A variety of sensors are known that use an electrochemical cell to provide output signals by which the presence or absence of an analyte, such as glucose, in a sample can be determined. For example, in an electrochemical cell, an analyte (or a species derived from it) that is electro-active generates a detectable signal at an electrode, and this signal can be used to detect or measure the presence and/or amount within a biological sample. In some conventional sensors, an enzyme is provided that reacts with the analyte to be measured, and the byproduct of the reaction is qualified or quantified at the electrode. An enzyme has the advantage that it can be very specific to an analyte and also, when the analyte itself is not sufficiently electro-active, can be used to interact with the analyte to generate another species which is electro-active and to which the sensor can produce a desired output. In one conventional amperometric glucose oxidase-based glucose sensor, immobilized glucose oxidase catalyses the oxidation of glucose to form hydrogen peroxide, which is then quantified by amperometric measurement (for example, change in electrical current) through a polarized electrode.

US 6,128,519 discloses a sensor assembly including a catheter having a catheter sheath adapted for insertion in a patient, and a catheter hub connected to the catheter sheath. An adapter is connected to the catheter hub, which includes first and second ports. A first flow line extends between the first port and a pump. A sensor is positioned along the first flow line. A second flow line extends between the second port and a container for holding a calibration fluid. The adapter defines first and second separate passages. The first passage extends from the first port to an interior region of the catheter sheath. The second passage extends from the second port to the interior region of the catheter hub.

**[0005]** US 4,736,748 discloses a blood component monitoring system for monitoring chemical substances of interest in blood, which includes a transfusion circuit including a flow-through cell adapted to be fluid-connected with a blood vessel through an indwelling catheter, a reservoir for accommodating a quantity of physiologically compatible sensor correcting solution, a connecting tubing extending between the reservoir and the flow-through cell and a transfusion pump; a detector unit including a temperature sensor and a chemical sensitive sensor device both incorporated inside the flow-through cell.

### SUMMARY OF THE INVENTION

**[0006]** In a first aspect, a method is provided for processing sensor data from a dual-electrode continuous analyte sensor configured for exposure to a circulatory system of a host *in vivo*, the method comprising: applying a dual-electrode continuous analyte sensor to a host, wherein the sensor comprises a first working electrode disposed beneath an enzymatic portion of a membrane system and a second working electrode disposed beneath a non-enzymatic portion of the membrane system, wherein the enzymatic portion comprises an enzyme for detecting an analyte and the non-enzymatic portion comprises no enzyme or an inactive form of the enzyme; receiving a first signal from the first working electrode associated with the analyte and non-analyte related electroactive compounds, and receiving a second signal from the second working electrode associated with the non-analyte related electroactive compounds, wherein the non-analyte related electroactive compounds have an oxidation potential that substantially overlaps with an oxidation potential of the analyte; estimating a scaling factor, wherein the scaling factor defines a relationship between the first working

electrode and the second working electrode; and processing the first signal and the second signal to obtain a signal substantially without contribution due to non-analyte related electroactive compounds, wherein the processing comprises using the scaling factor.

**[0007]** In an embodiment of the first aspect, the step of applying the sensor to a host comprises contacting the sensor with a fluid.

**[0008]** In an embodiment of the first aspect, the fluid is a bodily fluid and the step of estimating the scaling factor comprises comparing steady-state information of the first signal and steady-state information of the second signal.

**[0009]** In an embodiment of the first aspect, the fluid is a non-bodily fluid and the step of contacting comprises holding the non-bodily fluid substantially stagnant during a time period.

**[0010]** In an embodiment of the first aspect, the step of estimating the scaling factor comprises comparing a signal increase on each of the first working electrode and the second working electrode during the time period.

**[0011]** In an embodiment of the first aspect, the step of estimating a scaling factor comprises evaluating transient signal information for each of the first working electrode and the second working electrode.

**[0012]** In an embodiment of the first aspect, the step of estimating comprises determining a noise amplitude for each of the first working electrode and the second working electrode.

**[0013]** In an embodiment of the first aspect, the step of determining a noise amplitude comprises determining a signal residual for each of the first working electrode and the second working electrode.

**[0014]** In an embodiment of the first aspect, the step of determining a noise amplitude further comprises averaging a stream of signal residuals for each of the first working electrode and the second working electrode.

**[0015]** In an embodiment of the first aspect, the step of estimating is performed during a transient period of a signal, wherein the transient period of the signal comprises at least one of sensor break-in and signal artifact.

**[0016]** In a second aspect, a system for measuring an analyte is provided, comprising: a continuous analyte sensor configured for exposure to a circulatory system of a host *in vivo*, the continuous analyte sensor comprising a first working electrode disposed beneath an enzymatic portion of a membrane system and a second working electrode disposed beneath a non-enzymatic portion of the membrane system, wherein the enzymatic portion comprises an enzyme for detecting the analyte and the non-enzymatic portion comprises no enzyme or an inactive form of the enzyme; a vascular access device configured for fluid contact with a circulatory system of the host, wherein the sensor is located in or on the vascular access device; a receiving module configured to receive a first signal from the first working electrode and a second signal from the second working electrode, wherein the first signal is associated with the analyte and non-analyte related electroactive compounds, and the second signal is associated with the non-analyte related electroactive compounds, wherein the non-analyte related electroactive compounds have an oxidation potential that substantially overlaps with an oxidation potential of the analyte; and a processor module configured to process the first signal and the second signal and to estimate a scaling factor, wherein the scaling factor defines a relationship between the first working electrode and the second working electrode, and wherein the processor module is configured to process the first signal and the second signal using the scaling factor, whereby a signal substantially without contribution due to non-analyte related electroactive compounds is obtained.

**[0017]** In an embodiment of the second aspect, the system further comprises a flow control device configured to meter a flow of a fluid through the vascular access device.

**[0018]** In an embodiment of the second aspect, the fluid is a bodily fluid and the flow control device is configured to withdraw a sample of bodily fluid from the host, whereby the sensor is contacted with the bodily fluid.

**[0019]** In an embodiment of the second aspect, the fluid is a non-bodily fluid and the flow control device is configured to hold the non-bodily fluid substantially stagnant during a time period.

**[0020]** In an embodiment of the second aspect, the processor module is configured to compare a signal increase on each of the first working electrode and the second working electrode during the time period.

**[0021]** In an embodiment of the second aspect, the processor module is configured to evaluate transient signal information for each of the first working electrode and the second working electrode to estimate the scaling factor.

**[0022]** In an embodiment of the second aspect, the processor module is configured to determine a noise amplitude for each of the first working electrode and the second working electrode to estimate the scaling factor.

**[0023]** In an embodiment of the second aspect, the processor module is configured to determine the noise amplitude by determining a signal residual for each of the first working electrode and the second working electrode.

**[0024]** In an embodiment of the second aspect, the processor module is configured to average a stream of signal residuals to determine the noise amplitude for each of the first working electrode and the second working electrode.

**[0025]** In an embodiment of the second aspect, the processor module is configured to determine the noise amplitude during a transient period of a signal, wherein a transient period of the signal comprises at least one of sensor break-in and signal artifact.

**[0026]** In a third aspect, a system for measuring an analyte is provided, comprising: a continuous analyte sensor configured for continuous measurement of an analyte *in vivo,* comprising a first working electrode configured to generate a signal comprising analyte and non-analyte components and a second working electrode configured to generate a

second signal comprising a non-analyte related component; and a processor module configured to process the first signal and the second signal using a scaling factor, whereby a signal substantially without contribution due to the non-analyte component is obtained, wherein the scaling factor defines a relationship between the first working electrode and the second working electrode.

**[0027]** In an embodiment of the third aspect, the system further comprises a flow control device configured to expose the continuous analyte sensor to at least one fluid.

**[0028]** In an embodiment of the third aspect, the fluid is a sample of bodily fluid and the processor module is configured to process steady state information of the first signal and the second signal to estimate the scaling factor.

**[0029]** In an embodiment of the third aspect, the steady state information of the first signal and the second signal is generated after the analyte present in the sample has been substantially used up.

**[0030]** In an embodiment of the third aspect, the fluid is a non-bodily fluid and the processor module is configured to process the steady state information of the first signal and the second signal to estimate the scaling factor.

**[0031]** In an embodiment of the third aspect, the flow control device is configured to hold the non-bodily fluid substantially stagnant for a period of time, and wherein the processor module is configured to process the first signal and the second signal generated during a period of time to estimate the scaling factor.

**[0032]** In an embodiment of the third aspect, the flow control device is configured to wash the continuous analyte sensor with a non-bodily fluid for at least 50% of a time period during which the continuous analyte sensor is applied to a host.

**[0033]** In an embodiment of the third aspect, the flow control device is configured to wash the continuous analyte sensor with a non-bodily fluid for at least 80% of a time period during which the continuous analyte sensor is applied to a host.

**[0034]** In an embodiment of the third aspect, the scaling factor is determined *in vitro.*

**[0035]** In an embodiment of the third aspect, the scaling factor is at least one of automatically entered into the system, manually entered into the system, programmed into the system, and coded into the system.

**[0036]** In a fourth aspect, a continuous analyte detection system is provided, comprising: a continuous analyte sensor configured for contact with a sample from a circulatory system of a host and configured to generate a first signal and a second signal, wherein the first signal is associated with a test analyte and the second signal is associated with a reference analyte; a reference sensor configured to generate a reference signal associated with the reference analyte; and a processor module configured to process the second signal and the reference signal to calibrate the first signal.

**[0037]** In an embodiment of the fourth aspect, the continuous analyte sensor comprises a first working electrode and a second working electrode, wherein the first working electrode is disposed beneath an active enzymatic portion of a sensor membrane and is configured to generate a signal associated with analyte and non-analyte related electroactive compounds, wherein the non-analyte related electroactive compounds have an oxidation potential that substantially overlaps with an oxidation potential of the analyte; wherein the second working electrode is disposed beneath an inactive-enzymatic or a non-enzymatic portion of the sensor membrane and is configured to generate a signal associated with the non-analyte-related electroactive species, and wherein the processor module is configured to process signals from the first working electrode and second working electrode whereby a first signal substantially without a non-analyte-related signal component is generated.

**[0038]** In an embodiment of the fourth aspect, the first working electrode is further configured to generate the second signal.

**[0039]** In an embodiment of the fourth aspect, the second working electrode is further configured to generate the second signal.

**[0040]** In an embodiment of the fourth aspect, the continuous analyte sensor further comprises a third working electrode disposed beneath the sensor membrane and configured to generate the second signal.

**[0041]** In an embodiment of the fourth aspect, the reference sensor comprises an optical sensing apparatus.

**[0042]** In an embodiment of the fourth aspect, the reference sensor is disposed in a same local environment as the continuous analyte sensor.

**[0043]** In an embodiment of the fourth aspect, the continuous analyte sensor comprises a working electrode configured to generate both the first signal and the second signal.

**[0044]** In an embodiment of the fourth aspect, the system further comprises a flow control device configured to meter a flow of a fluid.

**[0045]** In a fifth aspect, a method for measuring an analyte in a host is provided, comprising: exposing a continuous analyte detection system to a sample, wherein the continuous analyte detection system comprises a continuous analyte sensor configured for contact with a sample from a circulatory system of a host *in vivo* and configured to generate a first signal associated with a test analyte and a second signal associated with a reference analyte, and a reference sensor configured to generate a reference signal associated with the reference analyte; receiving the first signal, the second signal, and the reference signal; calculating a calibration factor associated with a sensitivity of the continuous analyte sensor; and calibrating the first signal, wherein calibrating comprises using the calibration factor.

**[0046]** In an embodiment of the fifth aspect, the exposing step further comprises simultaneously exposing the continuous analyte sensor and the reference sensor to the sample.

**[0047]** In an embodiment of the fifth aspect, the receiving step further comprises receiving the first signal from a first working electrode disposed under an enzymatic portion of a membrane system.

**[0048]** In an embodiment of the fifth aspect, the receiving step further comprises receiving the second signal from the first working electrode.

**[0049]** In an embodiment of the fifth aspect, the receiving step further comprises receiving the second signal from a second working electrode disposed under the membrane system.

**[0050]** In an embodiment of the fifth aspect, the receiving step further comprises receiving a non-analyte-related signal from the second working electrode, wherein the second working electrode is disposed under a non-enzymatic portion of the membrane system.

**[0051]** In an embodiment of the fifth aspect, the receiving step further comprises receiving a non-analyte-related signal from a third working electrode disposed under a non-enzymatic portion of the membrane system.

**[0052]** In an embodiment of the fifth aspect, the receiving step further comprises optically detecting the reference analyte.

**[0053]** In an embodiment of the fifth aspect, the receiving step further comprises receiving a first signal associated with a glucose concentration of the sample.

**[0054]** In an embodiment of the fifth aspect, the receiving step further comprises receiving a second signal associated with an oxygen concentration of the sample, and a reference signal associated with the oxygen concentration of the sample.

**[0055]** In an embodiment of the fifth aspect, the exposing step comprises exposing the continuous analyte detection system to a bodily fluid and the calculating step further comprises comparing steady-state information of the first signal and steady-state information of the second signal.

**[0056]** In an embodiment of the fifth aspect, the exposing step comprises exposing the continuous analyte detection system to a substantially stagnant non-bodily fluid during a time period and the calculating step further comprises comparing a signal increase on each of the first and second working electrodes during the time period.

**[0057]** In a sixth aspect, a continuous analyte detection system is provided, comprising: an analyte sensor comprising a membrane system, wherein the analyte sensor is configured to generate a measurement signal associated with a measurement analyte concentration *in vivo,* and wherein the analyte sensor is further configured to generate a reference signal associated with a reference analyte concentration *in vivo*; a reference sensor located proximal to the analyte sensor and configured to generate a reference value associated with the reference analyte, wherein the reference sensor is located proximal to the analyte sensor; and a processor module configured to process the reference signal and the reference value to calibrate the measurement signal.

**[0058]** In an embodiment of the sixth aspect, the processor module is configured to calibrate the measurement signal without an external reference value.

**[0059]** In an embodiment of the sixth aspect, the system is configured for automatic calibration of the measurement signal.

**[0060]** In an embodiment of the sixth aspect, the system is configured such that the analyte sensor and the reference sensor are located within the same local environment such that the reference concentration measured by the analyte sensor and the reference concentration measured by the reference sensor are substantially equal.

**[0061]** In an embodiment of the sixth aspect, the analyte sensor is an electrochemical sensor and the reference sensor is an optical sensor.

**[0062]** In a seventh aspect, a continuous analyte sensor system is provided, comprising: a continuous analyte sensor configured for exposure to a circulatory system of a host and configured to generate a signal associated with an *in vivo* analyte concentration when the sensor is implanted in the host, and; sensor electronics configured to process the signal, wherein the sensor electronics comprise a fail-safe module configured to detect a malfunction of the system.

**[0063]** In an embodiment of the seventh aspect, the fail-safe module is further configured to detect an electrical malfunction.

**[0064]** In an embodiment of the seventh aspect, the electrical malfunction comprises a short circuit.

**[0065]** In an embodiment of the seventh aspect, the electrical malfunction is associated with at least one of start-up and sensor break-in.

**[0066]** In an embodiment of the seventh aspect, the fail-safe module is further configured to detect a fluidics malfunction.

**[0067]** In an embodiment of the seventh aspect, the system further comprises a flow control system in fluid communication with the sensor, wherein the system is configured to contact at least a portion of the sensor with a sample of the circulatory system, wherein the fluidics malfunction comprises a malfunction of the flow control system.

**[0068]** In an embodiment of the seventh aspect, the malfunction of the flow control system comprises at least one of a washing malfunction, a sample collection malfunction, a constriction of a component of the flow control system, and a blood clotting on a portion of the sensor.

**[0069]** In an embodiment of the seventh aspect, the flow control system comprises a vascular access device comprising a lumen, and at least a portion of the analyte sensor is further configured to reside within the lumen.

**[0070]** In an embodiment of the seventh aspect, the flow control system comprises a vascular access device and the analyte sensor is integrally formed with the vascular access device.

**[0071]** In an embodiment of the seventh aspect, the flow control system is configured to deliver a reference solution into the vascular access device.

**[0072]** In an embodiment of the seventh aspect, the fail-safe module is further configured to detect a sensor malfunction.

**[0073]** In an embodiment of the seventh aspect, the sensor malfunction comprises at least one of noise on the signal, drift of a sensitivity, drift of a baseline of the sensor, a broken component of the sensor, blood clotting on a portion of the sensor, and cross-talk.

**[0074]** In an embodiment of the seventh aspect, the fail-safe module is further configured to perform a waveform analysis of the signal.

**[0075]** In an embodiment of the seventh aspect, the fail-safe module is further configured to perform a steady state and/or transient state analysis of the signal.

**[0076]** In an embodiment of the seventh aspect, the fail-safe module is further configured to perform a steady state analysis and a transient analysis of the signal.

**[0077]** In an embodiment of the seventh aspect, the fail-safe module is further configured to evaluate a relationship between the steady state analysis and the transient analysis.

**[0078]** In an embodiment of the seventh aspect, the fail-safe module is further configured to provide at least one of an alert, an alarm and an instruction.

**[0079]** In an embodiment of the seventh aspect, the fail-safe module is further configured to evaluate a detected malfunction against a criterion.

**[0080]** In an embodiment of the seventh aspect, the analyte is glucose.

**[0081]** In an eighth aspect, a method for processing continuous analyte sensor data is provided, the method comprising: placing a continuous analyte sensor in fluid communication with a circulatory system of a host, wherein a sensor system comprises the sensor and sensor electronics, wherein the sensor is configured to generate a signal associated with an *in vivo* analyte concentration when the sensor is implanted in the host, and wherein the sensor electronics comprises a fail-safe module configured to detect a system malfunction; exposing the sensor to a sample from the host's circulatory system; and detecting a malfunction of the system.

**[0082]** In an embodiment of the eighth aspect, the method further comprises generating a signal associated with glucose.

**[0083]** In an embodiment of the eighth aspect, the detecting step further comprises detecting an electrical malfunction.

**[0084]** In an embodiment of the eighth aspect, the detecting step further comprises detecting a fluidics malfunction.

**[0085]** In an embodiment of the eighth aspect, the placing step further comprises inserting a vascular access device into the host's circulatory system.

**[0086]** In an embodiment of the eighth aspect, the placing step further comprises fluidly coupling the sensor to the vascular access device.

**[0087]** In an embodiment of the eighth aspect, the detecting step further comprises detecting a sensor malfunction.

**[0088]** In an embodiment of the eighth aspect, the detecting step further comprises performing a waveform analysis of the signal.

**[0089]** In an embodiment of the eighth aspect, the detecting step further comprises performing an equilibrium and/or kinetic analysis of the signal.

**[0090]** In an embodiment of the eighth aspect, the performing step further comprises evaluating a relationship between the equilibrium analysis and the kinetic analysis.

**[0091]** In an embodiment of the eighth aspect, the method further comprises providing at least one of an alert, an alarm, and an instruction.

**[0092]** In an embodiment of the eighth aspect, the method further comprises evaluating the detected malfunction against a criterion.

**[0093]** In an embodiment of the eighth aspect, the detecting step further comprises evaluating steady-state information and/or transient information.

**[0094]** In an embodiment of the eighth aspect, the evaluating step comprises evaluating at least one of sensitivity information and baseline information.

**[0095]** In a ninth aspect, a system is provided for continuously detecting an analyte in a host *in vivo*, comprising: a vascular access device configured for fluid communication with a circulatory system of a host; and a continuous analyte sensor, the sensor comprising a first working electrode disposed beneath an active enzymatic portion of a sensor membrane and configured to generate a first signal associated with associated with the analyte and non-analyte related electroactive compounds having a first oxidation potential, and a second working electrode disposed beneath an inactive-enzymatic or a non-enzymatic portion of the sensor membrane and configured to generate a second signal associated

with noise of the analyte sensor, wherein the noise comprises signal contribution due to non-analyte related electroactive species with an oxidation potential that substantially overlaps with the first oxidation potential.

**[0096]** In an embodiment of the ninth aspect, the first working electrode comprises a first electroactive surface and the second working electrode comprises a second electroactive surface, and wherein the first working electrode and the second working electrode are configured such that an area of the first electroactive surface exposed to a fluid is substantially equivalent to an area of the second electroactive surface exposed to a fluid.

**[0097]** In an embodiment of the ninth aspect, a configuration of the first working electrode and the second working electrode is at least one of bundled, twisted, and helical.

**[0098]** In an embodiment of the ninth aspect, the non-analyte related electroactive species comprise at least one species selected from the group consisting of interfering species, non-reaction-related hydrogen peroxide, and other electroactive species.

**[0099]** In an embodiment of the ninth aspect, the system further comprises electronics operably connected to the first working electrode and the second working electrode, and configured to process the first signal and the second signal to generate analyte concentration data substantially without signal contribution due to noise.

**[0100]** In an embodiment of the ninth aspect, the sensor comprises an electrical insulator located between the first working electrode and the second working electrode, wherein the insulator comprises a physical diffusion barrier configured to structurally block a substantial amount of diffusion of at least one of an analyte and a co-analyte between the first working electrode and the second working electrode by a structure that protrudes from a plane that intersects both the first working electrode and the second working electrode.

**[0101]** In an embodiment of the ninth aspect, the sensor comprises an insulator located between the first working electrode and the second working electrode, wherein the insulator comprises a diffusion barrier configured to substantially block diffusion of at least one of an analyte and a co-analyte between the first working electrode and the second working electrode wherein the diffusion barrier comprises a temporal diffusion barrier configured to block or avoid a substantial amount of diffusion or reaction of at least one of the analyte and the co-analyte between the first working electrode and the second working electrode.

**[0102]** In an embodiment of the ninth aspect, the sensor comprises an insulator located between the first working electrode and the second working electrode, wherein the insulator comprises a sensor membrane configured to substantially block diffusion of at least one of an analyte and a co-analyte between the first working electrode and the second working electrode by a discontinuity of the sensor membrane between the first working electrode and the second working electrode.

**[0103]** In an embodiment of the ninth aspect, the first working electrode and the second working electrode are spaced a distance greater than a diffusion distance of at least one of an analyte and a co-analyte such that cross-talk substantially does not occur.

**[0104]** In an embodiment of the ninth aspect, the first working electrode and the second working electrode are configured and arranged around a circumference of the sensor.

**[0105]** In an embodiment of the ninth aspect, the vascular access device comprises a lumen and at least a portion of the sensor is disposed within the lumen.

**[0106]** In an embodiment of the ninth aspect, the vascular access device comprises a hub and the continuous analyte sensor is disposed substantially within the hub.

**[0107]** In an embodiment of the ninth aspect, the sensor is configured to reside substantially above a plane defined by the host's skin.

**[0108]** In an embodiment of the ninth aspect, the sensor is disposed on a surface of the vascular access device.

**[0109]** In an embodiment of the ninth aspect, the vascular access device is configured for insertion into at least one of an artery, a vein, a fistula, and an extracorporeal circulatory device configured to circulate at least a portion of the host's blood outside of the host's body.

**[0110]** In an embodiment of the ninth aspect, the system further comprises a flow control device configured to meter a flow of a fluid through the vascular access device.

**[0111]** In an embodiment of the ninth aspect, the flow control device is configured to meter a flow of a sufficient flow rate of a non-bodily fluid such that the sensor contacts the non-bodily fluid for a sufficient amount of time, such that biofouling does not occur for at least about 3 days of sensor use.

**[0112]** In an embodiment of the ninth aspect, the sufficient amount of time comprises at least about 50% of a sensor session.

**[0113]** In an embodiment of the ninth aspect, the flow control device is configured to control fluid contact with the continuous analyte sensor.

**[0114]** In an embodiment of the ninth aspect, the flow control device meters the non-bodily fluid through the vascular access device for a sufficient amount of time with a sufficient flow rate such that the vascular access device remains patent during a sensor session.

**[0115]** In an embodiment of the ninth aspect, the sufficient amount of time comprises at least about 50% of a sensor

session.

**[0116]** In an embodiment of the ninth aspect, the system further comprises an electronics module configured to determine a scaling factor that defines a relationship between the first working electrode and the second working electrode.

**[0117]** In an embodiment of the ninth aspect, the system further comprises a fluid coupler configured and arranged to mate with a vascular access device on a first end, and wherein the sensor is at least one of disposed within at least one of a portion of the fluid coupler and disposed at a surface of the fluid coupler.

**[0118]** In an embodiment of the ninth aspect, the system is configured to calibrate the continuous analyte sensor using a reference fluid.

**[0119]** In an embodiment of the ninth aspect, the system is configured to auto calibrate without an external reference value.

**[0120]** In an embodiment of the ninth aspect, the system is configured to calibrate the sensor without a reference data point provided by an external analyte monitor.

**[0121]** In an embodiment of the ninth aspect, the system is configured to calibrate the sensor using single-point calibration.

**[0122]** In an embodiment of the ninth aspect, the system further comprises a reference sensor configured to generate a reference signal associated with a reference analyte in the sample, wherein the continuous analyte sensor is further configured to generate a third signal associated with the reference analyte, and wherein the system is configured to calibrate the continuous analyte sensor using the reference signal and the third signal.

**[0123]** In an embodiment of the ninth aspect, the reference sensor comprises an optical sensing apparatus.

**[0124]** In an embodiment of the ninth aspect, the reference sensor and the continuous analyte sensor are configured for simultaneous exposure to a sample of the circulatory system.

**[0125]** In an embodiment of the ninth aspect, the continuous analyte sensor is a glucose sensor.

**[0126]** In an embodiment of the ninth aspect, a substantial portion of the continuous analyte sensor has a diameter of less than about 0.025 inches.

**[0127]** In an embodiment of the ninth aspect, the continuous analyte sensor further comprises a bioinert material or a bioactive agent incorporated therein or thereon.

**[0128]** In an embodiment of the ninth aspect, the bioactive agent comprises at least one agent selected from the group consisting of vitamin K antagonists, heparin group anticoagulants, platelet aggregation inhibitors, enzymes, direct thrombin inhibitors, Dabigatran, Defibrotide, Dermatan sulfate, Fondaparinux, and Rivaroxaban.

**[0129]** In a tenth aspect, a method is provided for continuously detecting an analyte in the host *in vivo,* comprising: inserting a vascular access device into a circulatory system of a host; contacting a continuous analyte sensor with a sample from the circulatory system; generating a first signal associated with the analyte and non-analyte related electroactive compounds having a first oxidation potential in the sample; generating a second signal associated with noise of the analyte sensor, wherein the noise comprises signal contribution due to non-analyte related electroactive species with an oxidation potential that substantially overlaps with the first oxidation potential in the sample; and processing the first signal and the second signal to provide a processed signal substantially without a signal component associated with noise.

**[0130]** In an embodiment of the tenth aspect, the method further comprises contacting the continuous analyte sensor with a reference solution, whereby at least one reference data point is provided.

**[0131]** In an embodiment of the tenth aspect, the method further comprises auto calibrating the continuous analyte sensor using the reference data point.

**[0132]** In an embodiment of the tenth aspect, the auto calibrating comprises repeatedly contact the continuous analyte sensor with the reference solution during a sensor session.

**[0133]** In an embodiment of the tenth aspect, the contacting step comprises withdrawing a blood sample.

**[0134]** In an embodiment of the tenth aspect, the processing step further comprises determining a scaling factor that defines a relationship between the first working electrode and the second working electrode.

**[0135]** In an embodiment of the tenth aspect, the processing step further comprises calibrating the continuous analyte sensor using the scaling factor.

**[0136]** In an embodiment of the tenth aspect, the method further comprises contacting a reference sensor with the sample.

**[0137]** In an embodiment of the tenth aspect, the method further comprises generating a third signal associated with a reference analyte in the sample.

**[0138]** In an embodiment of the tenth aspect, the method further comprises optically generating a reference signal associated with the reference sensor.

**[0139]** In an embodiment of the tenth aspect, the method further comprises calibrating the processed signal using the third signal and the reference signal.

**[0140]** In an embodiment of the tenth aspect, the analyte is glucose.

**[0141]** In an embodiment of the tenth aspect, the processing step comprises evaluating steady-state information and

transient information, wherein the first and second signals each comprise steady state and transient information.

**[0142]** In an embodiment of the tenth aspect, the evaluating step comprises evaluating at least one of sensitivity information and baseline information.

**[0143]** In an eleventh aspect, a method is provided for continuously measuring an analyte in an artery of a host *in vivo,* the method comprising: coupling a continuous analyte sensor with an arterial catheter system applied to a host, wherein the sensor is configured to generate an analyte-related signal associated with an analyte in a sample, and wherein the arterial catheter system comprises an arterial catheter, an infusion fluid, and a pressure system configured to perform at least one of increasing an amount of pressure applied to the infusion fluid and reducing an amount of pressure applied to the infusion fluid; reducing the amount of pressure applied to the infusion fluid, such that a sample of arterial blood contacts the sensor; and generating the analyte-related signal with the sensor.

**[0144]** In an embodiment of the eleventh aspect, the method further comprises reinfusing the sample into the host.

**[0145]** In an embodiment of the eleventh aspect, the reinfusing step comprises increasing the amount of pressure applied to the infusion fluid.

**[0146]** In an embodiment of the eleventh aspect, the generating step further comprises generating a second signal with the sensor, wherein sensor comprises a first working electrode configured to generate a first signal comprising an analyte-related signal component and a non-analyte-related signal component and the second working electrode is configured to generate the second signal comprising the non-analyte-related signal component.

**[0147]** In an embodiment of the eleventh aspect, the method further comprises processing the first signal and second signal to provide a processed signal substantially without a signal component due to the non-analyte-related signal component.

**[0148]** In an embodiment of the eleventh aspect, the method further comprises processing the first signal and second signal to provide a scaling factor.

**[0149]** In an embodiment of the eleventh aspect, the method further comprises monitoring an arterial blood pressure of the host using a pressure transducer.

**[0150]** In an embodiment of the eleventh aspect, the coupling step comprises coupling the sensor to the arterial catheter.

**[0151]** In an embodiment of the eleventh aspect, the coupling step comprises inserting the sensor into a lumen of the catheter, wherein the catheter comprises at least one lumen.

**[0152]** In an embodiment of the eleventh aspect, the generating step further comprises generating a reference signal associated with a reference analyte in the sample, wherein the sensor further comprises a reference sensor configured to generate the reference signal.

**[0153]** In an embodiment of the eleventh aspect, the method further comprises processing the analyte-related signal to provide an analyte value, wherein the sensor further comprises a processor module configured to process the signal.

**[0154]** In an embodiment of the eleventh aspect, the method further comprises calibrating the signal.

**[0155]** In a twelfth aspect, a system is provided for continuously measuring an analyte in an artery of a host *in vivo*, the system comprising: an arterial infusion system configured and arranged to meter at least one of flow of a fluid into an artery of a host and flow of a fluid out of an artery of a host, the arterial infusion system comprising an arterial catheter, an infusion fluid, and a pressure system configured to perform at least one of increasing an amount of pressure applied to the infusion fluid and reducing an amount of pressure applied to the infusion fluid, wherein when the infusion system is applied to the host, the pressure system is further configured to infuse the infusion fluid, withdraw a blood sample, and reinfuse the withdrawn blood sample into the host; and a continuous analyte sensor configured to couple with the arterial infusion system, configured to contact a sample of the host, and configured to generate a first signal associated with an analyte in the sample.

**[0156]** In an embodiment of the twelfth aspect, the sensor comprises a first working electrode and a second working electrode, wherein the first working electrode is configured to generate the first signal comprising an analyte-related signal component and a non-analyte-related signal component, and wherein the second working electrode is configured to generate a second signal comprising the non-analyte related signal component.

**[0157]** In an embodiment of the twelfth aspect, the system further comprises a processor module configured to process the first signal and second signal to provide a scaling factor.

**[0158]** In an embodiment of the twelfth aspect, the processor module is further configured to calibrate the first signal using the scaling factor.

**[0159]** In an embodiment of the twelfth aspect, the system further comprises a reference sensor configured to generate are reference signal associated with a reference analyte in the sample.

**[0160]** In an embodiment of the twelfth aspect, the processor module is further configured to calibrate the first signal using the scaling factor.

**[0161]** In an embodiment of the twelfth aspect, the system further comprises a processor module configured to process the first signal.

**[0162]** In an embodiment of the twelfth aspect, the system further comprises a processor module configured to calibrate the sensor.

**[0163]** In an embodiment of the twelfth aspect, the system further comprises electronics configured and arranged to regulate the pressure system.

**[0164]** In a thirteenth aspect, a system for continuous measurement of a glucose concentration is provided, the system comprising: a continuous glucose sensor configured to generate a signal associated with an *in vivo* glucose concentration in a host's circulatory system; and a flow control device configured to intermittently meter a reference solution across the continuous glucose sensor.

**[0165]** In an embodiment of the thirteenth aspect, the signal does not substantially comprise a baseline component.

**[0166]** In an embodiment of the thirteenth aspect, the continuous glucose sensor comprises a first working electrode and a second working electrode, wherein the system is configured to process signals received from the first working electrode and second working electrode to provide the signal substantially without a baseline component.

**[0167]** In an embodiment of the thirteenth aspect, the flow control device is configured such that the continuous glucose sensor intermittently measures a glucose concentration of the reference solution.

**[0168]** In an embodiment of the thirteenth aspect, the system is configured to automatically calibrate the continuous glucose sensor using the measured glucose concentration of the reference solution.

**[0169]** In an embodiment of the thirteenth aspect, the signal does not substantially comprise a baseline component.

**[0170]** In an embodiment of the thirteenth aspect, the flow control device is further configured to intermittently meter a blood sample from the host's circulatory system across the continuous glucose sensor.

**[0171]** In an embodiment of the thirteenth aspect, the flow control device is configured to meter the reference solution such that the sensor contacts the reference solution at least about 50% of the time during a sensor session.

**[0172]** In an embodiment of the thirteenth aspect, the reference solution comprises glucose.

**[0173]** In an embodiment of the thirteenth aspect, the flow control device is configured to meter the reference solution such that the sensor is contacting the reference solution a sufficient amount of time such that biofouling does not occur for a sensor session of at least about 3 days.

**[0174]** In an embodiment of the thirteenth aspect, the system further comprises a vascular access device, wherein the sensor is located at at least one of in the vascular access device and on the vascular access device, and wherein the flow control device is configured to meter the reference solution through the vascular access device for a sufficient amount of time with a sufficient flow rate such that the vascular access device remains patent during a sensor session of at least about 3 days.

**[0175]** In a fourteenth aspect, a system is provided for continuous measurement of a glucose concentration, the system comprising: a continuous glucose sensor comprising a first working electrode and a second working electrode, wherein the continuous glucose sensor is located at at least one of in a vascular access device and on a vascular access device in fluid communication with a host's circulatory system; and a flow control device configured to intermittently meter a glucose reference solution across the continuous glucose sensor such that the continuous glucose sensor intermittently measures the glucose concentration of the reference solution, and wherein the system is configured to calibrate the continuous glucose sensor using the measured glucose concentration of the reference solution.

**[0176]** In an embodiment of the fourteenth aspect, the system is configured to process signals from the first working electrode and second working electrode to obtain a substantially baseline-free glucose signal.

**[0177]** In an embodiment of the fourteenth aspect, the system is configured to calibrate the continuous glucose sensor using an equation that does not include a baseline parameter.

**[0178]** In an embodiment of the fourteenth aspect, the system is configured to use the measured glucose concentration of the reference solution to determine a sensitivity of the continuous glucose sensor.

**[0179]** In an embodiment of the fourteenth aspect, the system is configured to calibrate the continuous glucose sensor using the sensitivity of the continuous glucose sensor.

**[0180]** In an embodiment of the fourteenth aspect, the system is configured to auto-calibrate the sensor without an external reference value.

**[0181]** In a fifteenth aspect, a system for continuous measurement of an analyte concentration is provided, the system comprising: a continuous analyte sensor located at at least one of in a vascular access device and on a vascular access device in fluid communication with a host's circulatory system; and a flow control device configured to intermittently meter a reference solution across the continuous analyte sensor such that the continuous analyte sensor is in continuous contact with either the reference solution or a blood sample from the host's circulatory system, wherein the flow control device is configured such that the continuous glucose sensor is in contact with the glucose reference solution at least about 50% of the time during a sensor session.

**[0182]** In an embodiment of the fifteenth aspect, the flow control device is configured to meter the reference solution such that the sensor is contacting the reference solution at least about 65% of the time during a sensor session.

**[0183]** In an embodiment of the fifteenth aspect, the flow control device is configured to meter the reference solution such that the sensor is contacting the reference solution at least about 80% of the time during a sensor session.

**[0184]** In an embodiment of the fifteenth aspect, the flow control device is configured to meter the reference solution such that the sensor is contacting the reference solution a sufficient amount of time such that biofouling does not occur

for at least about 3 days of sensor use.

**[0185]** In an embodiment of the fifteenth aspect, the flow control device is configured to meter reference solution such that the sensor is contacting the reference solution a sufficient amount of time such that biofouling does not occur for at least about 7 days of sensor use.

**[0186]** In an embodiment of the fifteenth aspect, the flow control device is configured to meter reference solution such that the sensor is contacting the reference solution a sufficient amount of time such that biofouling does not occur for at least about 21 days of sensor use.

**[0187]** In an embodiment of the fifteenth aspect, the flow control device is configured to meter reference solution such that the sensor is contacting the reference solution a sufficient amount of time such that biofouling does not occur for at least about 30 days of sensor use.

**[0188]** In an embodiment of the fifteenth aspect, the sufficient amount of time is at least about 50% of the time during sensor use.

**[0189]** In an embodiment of the fifteenth aspect, the sufficient amount of time is at least about 65% of the time during sensor use.

**[0190]** In an embodiment of the fifteenth aspect, the sufficient amount of time is at least about 80% of the time during sensor use.

**[0191]** In an embodiment of the fifteenth aspect, the flow control device is configured to meter the reference solution through the vascular access device for a sufficient amount of time with a sufficient flow rate such that the vascular access device remains patent during a sensor session of at least about 3 days.

**[0192]** In an embodiment of the fifteenth aspect, the flow control device is configured to meter the reference solution through the vascular access device for at least about 50% of a sensor session, at a flow rate from about 0.001 ml/min to about 2.0 ml/min.

**[0193]** In an embodiment of the fifteenth aspect, the flow control device is configured to meter the reference solution through the vascular access device for at least about 65% of a sensor session, at a flow rate from about 0.5 ml/min to about 2.0 ml/min, whereby the vascular access device remains patent during a sensor session of from about 3 days to about 30 days.

**[0194]** In a sixteenth aspect, a device is provided for the detection of at least one analyte in a circulatory system of a host *in vivo,* comprising: an apparatus configured for fluid communication with a circulatory system of a host, wherein the apparatus comprises a lumen, an external surface, a first orifice and a second orifice, wherein at least one of the first orifice and the second orifice is configured to couple with a fluid flow device; and a plurality of sensors disposed within the lumen of the apparatus.

**[0195]** In an embodiment of the sixteenth aspect, the apparatus further comprises a plurality of sensor sites, wherein each sensor site is configured to receive a sensor.

**[0196]** In an embodiment of the sixteenth aspect, at least one of the sensor sites comprises a breakaway portion configured for insertion of a sensor therethrough, whereby at least a portion of the sensor is disposed within the lumen.

**[0197]** In an embodiment of the sixteenth aspect, at least another portion of the sensor is disposed at the external surface.

**[0198]** In an embodiment of the sixteenth aspect, the apparatus is a vascular access device comprising an *in vivo* portion and an *ex vivo* portion, and wherein the plurality of sensors are disposed within the *ex vivo* portion.

**[0199]** In an embodiment of the sixteenth aspect, the apparatus is configured to be disposed outside of the host's body.

**[0200]** In an embodiment of the sixteenth aspect, at least one of the sensors is configured to generate a signal associated with a concentration of an analyte in a sample from the host's circulatory system.

**[0201]** In an embodiment of the sixteenth aspect, at least two of the sensors are configured to generate signals associated with a concentration at least one analyte.

**[0202]** In an embodiment of the sixteenth aspect, the two sensors are configured to generate signals associated with a concentration of the analyte.

**[0203]** In an embodiment of the sixteenth aspect, the analyte is selected from the group consisting of glucose, oxygen, lactate, glutamine, succinate, Cytochrome Oxidase, a medicament, and heparin.

**[0204]** In an embodiment of the sixteenth aspect, at least one of the sensors is configured to generate a signal associated with a property of a sample from the host's circulatory system.

**[0205]** In an embodiment of the sixteenth aspect, the property is selected from the group consisting of pH, temperature, pressure, hematocrit, and oxygen tension.

**[0206]** In an embodiment of the sixteenth aspect, the sensors are disposed above a plane defined by the host's skin.

**[0207]** In an embodiment of the sixteenth aspect, the sensors are integrally formed within the apparatus.

**[0208]** In an embodiment of the sixteenth aspect, the apparatus is a vascular access device comprising an *in vivo* portion and an *ex vivo* portion, and wherein at least one of the sensors is disposed within the *in vivo* portion.

**[0209]** In an embodiment of the sixteenth aspect, at least one of the sensors is deposited within the lumen.

**[0210]** In an embodiment of the sixteenth aspect, at least one of the sensors is screen-printed within the lumen.

**[0211]** In an embodiment of the sixteenth aspect, the apparatus is injection molded around at least one of the plurality of sensors.

**[0212]** In an embodiment of the sixteenth aspect, at least one of the sensors is received within the lumen.

**[0213]** In a seventeenth aspect, a method is provided for making a device for the detection of a plurality of analytes in a sample from a circulatory system of a host *in vivo,* the method comprising: providing a plurality of sensors; and forming an apparatus about the plurality of sensors, wherein the apparatus comprises a lumen, an external surface, and at least one orifice configured for coupling with a fluid flow device.

**[0214]** In an eighteenth aspect, a method is provided making a device for the detection of a plurality of analytes in a sample from a circulatory system of a host *in vivo*, the method comprising: providing an apparatus comprising a lumen, an external surface, and at least one orifice configured for coupling with a fluid flow device a plurality of sensors; and forming a plurality of sensors situated at at least one of within the apparatus and on the apparatus.

**[0215]** In a nineteenth aspect, a method is provided for detecting of a plurality of analytes in a sample from a circulatory system of a host *in vivo,* the method comprising: applying an apparatus to a circulatory system of a host, the apparatus comprising a lumen and a plurality of sensors, wherein the at least two sensors are disposed above a plane defined by the skin of the host; withdrawing a sample from the circulatory system of the host; contacting the plurality of sensors with the sample; and generating a signal from each of the sensors.

**[0216]** In an embodiment of the nineteenth aspect, the methd further comprises processing the signals from each of the sensors.

**[0217]** In an embodiment of the nineteenth aspect, the generating step comprises at least one of electrochemically generating, optically generating, radiochemically generating, physically generating, chemically generating, immunochemically generating, and/ enzymatically generating a signal from each of the plurality of sensors.

**[0218]** In an embodiment of the nineteenth aspect, the methd further comprises reinfusing the withdrawn sample into the host.

**[0219]** In an embodiment of the nineteenth aspect, the methd further comprises washing the sensors with an infusion fluid.

**[0220]** In an embodiment of the nineteenth aspect, the methd further comprises calibrating the signal of at least one of the sensors.

## BRIEF DESCRIPTION OF THE DRAWINGS

**[0221]**

Fig. 1A is a perspective view of one embodiment of an analyte sensor system, including a vascular access device (e.g., a catheter), a sensor, a fluid connector, and a protective sheath.

Fig. 1B is a side view of the analyte sensor system of Fig. 1A, showing the protective sheath removed.

Fig. $1C_1$ is a close-up cut away view of a portion of the analyte sensor system of Fig. 1A.

Fig. $1C_2$ is a close-up cut away view of a portion of the analyte sensor system of Fig. 1A.

Fig. 1D is a close-up cut away view of a portion of the analyte sensor system of Fig. 1A.

Fig. 1E is a close-up cut away view of a portion of the analyte sensor system of Fig. 1A.

Fig. 2A is a perspective view of another embodiment of the analyte sensor system, including a catheter with a sensor integrally formed thereon.

Fig. 2B is a perspective view of the analyte sensor system of Fig. 2A.

Fig. 2C is a close-up view of a portion of the analyte sensor system of Fig. 2A in an alternative configuration of an embodiment having three electrodes disposed on the catheter.

Fig. 2D is a close-up view of a portion of the analyte sensor system of Fig. 2A in an alternative configuration of an embodiment having three electrodes disposed on the catheter.

Fig. 2E is a close-up view of a portion of the analyte sensor system of Fig. 2A in an alternative embodiment having two electrodes disposed on the catheter.

Fig. 2F is a close-up view of a portion of the analyte sensor system of Fig. 2A in an alternative embodiment having one electrode disposed on the catheter.

Fig. 2G is a cross-section of analyte sensor system in one embodiment, including a plurality of analyte sensors disposed within the connector of a catheter.

Fig. 2H is a cross-section of analyte sensor system in one embodiment, including a plurality of analyte sensors disposed within a fluid coupler, such as but not limited to a connector, a valve, and a Leur lock.

Fig. 2I is a cross-section of analyte sensor system of Fig. 2H, taken along line 2I - 2I.

Fig. 2J is a cross-section of analyte sensor system of Fig. 2H, taken along line 2I - 2I.

Fig. 2K is a cross-section of analyte sensor system of Fig. 2H, taken along line 2I - 2I.

Fig. 2L is a cross-section of analyte sensor system of Fig. 2H, taken along line 2I - 2I.

Fig. 3A is a perspective view of a first portion of one embodiment of an analyte sensor.

**Fig. 3B** is a perspective view of a second portion of the analyte sensor of Fig. 3A.

Fig. 3C is a cross section of the analyte sensor of **Fig. 3B,** taken on line C -C.

Fig. 3D is a cross-sectional schematic view of a sensing region of a dual-electrode continuous analyte sensor in one embodiment wherein an active enzyme of an enzyme domain is positioned over the first working electrode but not over the second working electrode.

Fig. 3E is a perspective view of a dual-electrode continuous analyte sensor in one embodiment.

Fig. 3F is a schematic illustrating metabolism of glucose by Glucose Oxidase (GOx) and one embodiment of a diffusion barrier D that substantially prevents the diffusion of $H_2O_2$ produced on a first side of the sensor (e.g., from a first working electrode that has active GOx) to a second side of the sensor (e.g., to the second working electrode that lacks active GOx).

Fig. 3G is a two-dimensional schematic of a dual-electrode sensor in one embodiment, illustrating the sensor's first and second electroactive surfaces (of the first and second working electrodes, respectively) beneath a sensor membrane, wherein noise-causing species produced by a plurality of point sources can impinge upon an electroactive surface.

Fig. 3H is a two-dimensional schematic of a dual-electrode sensor in one embodiment, illustrating the sensor's first and second electroactive surfaces (of the first and second working electrodes, respectively) beneath a sensor membrane, wherein noise from a single point source (e.g., a cell) can impinge upon both electroactive surfaces.

Fig. 3I is a cross-sectional schematic illustrating a dual-electrode sensor, in one embodiment, including a physical diffusion barrier.

Fig. 3J is a graph illustrating signal response of the electrodes of a dual-electrode sensor, in one embodiment.

Fig. 3K is a graph illustrating signal response of the electrodes of a dual-electrode sensor, in one embodiment.

Fig. 4 is a graph illustrating *in vivo* function of an analyte sensor system of the embodiment shown in Fig. 1A.

Fig. 5 is a graph illustrating *in vivo* function of an analyte sensor system of the embodiment shown in Fig. 1A.

Fig. 6 is a schematic of an integrated sensor system.

Fig. 7 is a block diagram of an integrated sensor system

Figs. 8A through 8C are schematic illustrations of a flow control device in one exemplary embodiment, including is relative movement/positions and the consequential effect on the flow of fluids through the sensor/catheter inserted in a host.

Fig. 9 is a cut-away illustration of one exemplary embodiment of a catheter implanted in a host's vessel.

Fig. 10 is a graph that schematically illustrates a signal produced during exposure of the sensor to a step change in analyte concentration, in one exemplary embodiment.

Fig. 11 is a graph that schematically illustrates a derivative of the step response shown in Fig. 9.

Fig. 12 is a graph that illustrates level vs. rate for a plurality of time-spaced signals associated with exposure of the sensor to biological samples of unknown or uncalibrated analyte concentration.

Fig. 13 is a graphical representation showing exemplary glucose sensor data and corresponding blood glucose values over time in a pig.

Fig. 14 is a graphical representation showing exemplary calibrated glucose sensor data (test) and corresponding blood glucose values (YSI control) over time in a human.

Fig. 15A is a graph that illustrates an *in vivo* signal (counts) detected from a dual-electrode sensor, in one embodiment, implanted in a non-diabetic human host.

Fig. 15B is a graph that illustrates an *in vivo* signal (counts) detected from a dual-electrode, in another embodiment, implanted in a non-diabetic human host.

Fig. 16A is a graph that illustrates an *in vivo* signal (counts) detected from a dual-electrode, in one embodiment, implanted in a non-diabetic human host.

Fig. 16B is a graph that illustrates an *in vivo* signal (counts) detected from a dual-electrode, in another embodiment, implanted in a non-diabetic human host.

Fig. 17A is a graph that illustrates an *in vivo* glucose values detected from a dual-electrode, in another embodiment, implanted in a non-diabetic porcine host.

Fig. 17B is a Clark Error Grid graph of the data of Fig. 17A.

## DETAILED DESCRIPTION OF THE PREFERRED EMBODIMENT

**[0222]** The following description and examples illustrate some exemplary embodiments of the disclosed invention in detail. Those of skill in the art will recognize that there are numerous variations and modifications of this invention that are encompassed by its scope. Accordingly, the description of a certain exemplary embodiment should not be deemed to limit the scope of the preferred embodiments.

Definitions

**[0223]** In order to facilitate an understanding of the preferred embodiments, a number of terms are defined below.

**[0224]** The term "analyte" as used herein is a broad term, and is to be given its ordinary and customary meaning to a person of ordinary skill in the art (and is not to be limited to a special or customized meaning), and refers without limitation to a substance or chemical constituent in a biological fluid (for example, blood, interstitial fluid, cerebral spinal fluid, lymph fluid or urine) that can be analyzed. Analytes can include naturally occurring substances, artificial substances, metabolites, and/or reaction products. In some embodiments, the analyte for measurement by the sensing regions, devices, and methods is glucose. However, other analytes are contemplated as well, including but not limited to acarboxyprothrombin; acylcarnitine; adenine phosphoribosyl transferase; adenosine deaminase; albumin; alpha-fetoprotein; amino acid profiles (arginine (Krebs cycle), histidine/urocanic acid, homocysteine, phenylalanine/tyrosine, tryptophan); andrenostenedione; antipyrine; arabinitol enantiomers; arginase; benzoylecgonine (cocaine); biotinidase; biopterin; c-reactive protein; carnitine; carnosinase; CD4; ceruloplasmin; chenodeoxycholic acid; chloroquine; cholesterol; cholinesterase; conjugated 1-ß hydroxy-cholic acid; cortisol; creatine kinase; creatine kinase MM isoenzyme; cyclosporin A; d-penicillamine; de-ethylchloroquine; dehydroepiandrosterone sulfate; DNA (acetylator polymorphism, alcohol dehydrogenase, alpha 1-antitrypsin, cystic fibrosis, Duchenne/Becker muscular dystrophy, glucose-6-phosphate dehydrogenase, hemoglobin A, hemoglobin S, hemoglobin C, hemoglobin D, hemoglobin E, hemoglobin F, D-Punjab, beta-thalassemia, hepatitis B virus, HCMV, HIV-1, HTLV-1, Leber hereditary optic neuropathy, MCAD, RNA, PKU, Plasmodium vivax, sexual differentiation, 21-deoxycortisol); desbutylhalofantrine; dihydropteridine reductase; diptheria/tetanus antitoxin; erythrocyte arginase; erythrocyte protoporphyrin; esterase D; fatty acids/acylglycines; free ß-human chorionic gonadotropin; free erythrocyte porphyrin; free thyroxine (FT4); free tri-iodothyronine (FT3); fumarylacetoacetase; galactose/gal-1-phosphate; galactose-1-phosphate uridyltransferase; gentamicin; glucose-6-phosphate dehydrogenase; glutathione; glutathione perioxidase; glycocholic acid; glycosylated hemoglobin; halofantrine; hemoglobin variants; hexosaminidase A; human erythrocyte carbonic anhydrase I; 17-alpha-hydroxyprogesterone; hypoxanthine phosphoribosyl transferase; immunoreactive trypsin; lactate; lead; lipoproteins ((a), B/A-1, ß); lysozyme; mefloquine; netilmicin; phenobarbitone; phenytoin; phytanic/pristanic acid; progesterone; prolactin; prolidase; purine nucleoside phosphorylase; quinine; reverse tri-iodothyronine (rT3); selenium; serum pancreatic lipase; sissomicin; somatomedin C; specific antibodies (adenovirus, anti-nuclear antibody, anti-zeta antibody, arbovirus, Aujeszky's disease virus, dengue virus, Dracunculus medinensis, Echinococcus granulosus, Entamoeba histolytica, enterovirus, Giardia duodenalisa, Helicobacter pylori, hepatitis B virus, herpes virus, HIV-1, IgE (atopic disease), influenza virus, Leishmania donovani, leptospira, measles/mumps/rubella, Mycobacterium leprae, Mycoplasma pneumoniae, Myoglobin, Onchocerca volvulus, parainfluenza virus, Plasmodium falciparum, poliovirus, Pseudomonas aeruginosa, respiratory syncytial virus, rickettsia (scrub typhus), Schistosoma mansoni, Toxoplasma gondii, Trepenoma pallidium, Trypanosoma cruzi/rangeli, vesicular stomatis virus, Wuchereria bancrofti, yellow fever virus); specific antigens (hepatitis B virus, HIV-1); succinylacetone; sulfadoxine; theophylline; thyrotropin (TSH); thyroxine (T4); thyroxine-binding globulin; trace elements; transferrin; UDP-galactose-4-epimerase; urea; uroporphyrinogen I synthase; vitamin A; white blood cells; and zinc protoporphyrin. Salts, sugar, protein, fat, vitamins, and hormones naturally occurring in blood or interstitial fluids can also constitute analytes in certain embodiments. The analyte can be naturally present in the biological fluid, for example, a metabolic product, a hormone, an antigen, an antibody, and the like. Alternatively, the analyte can be introduced into the body, for example, a contrast agent for imaging, a radioisotope, a chemical agent, a fluorocarbon-based synthetic blood, or a drug or pharmaceutical composition, including but not limited to insulin; ethanol; cannabis (marijuana, tetrahydrocannabinol, hashish); inhalants (nitrous oxide, amyl nitrite, butyl nitrite, chlorohydrocarbons, hydrocarbons); cocaine (crack cocaine); stimulants (amphetamines, methamphetamines, Ritalin, Cylert, Preludin, Didrex, PreState, Voranil, Sandrex, Plegine); depressants (barbituates, methaqualone, tranquilizers such as Valium, Librium, Miltown, Serax, Equanil, Tranxene); hallucinogens (phencyclidine, lysergic acid, mescaline, peyote, psilocybin); narcotics (heroin, codeine, morphine, opium, meperidine, Percocet, Percodan, Tussionex, Fentanyl, Darvon, Talwin, Lomotil); designer drugs (analogs of fentanyl, meperidine, amphetamines, methamphetamines, and phencyclidine, for example, Ecstasy); anabolic steroids; and nicotine. The metabolic products of drugs and pharmaceutical compositions are also contemplated analytes. Analytes such as neurochemicals and other chemicals generated within the body can also be analyzed, such as, for example, ascorbic acid, uric acid, dopamine, noradrenaline, 3-methoxytyramine (3MT), 3,4-dihydroxyphenylacetic acid (DOPAC), homovanillic acid (HVA), 5-hydroxytryptamine (5HT), histamine, Advanced Glycation End Products (AGEs) and 5-hydroxyindoleacetic acid (FHIAA).

**[0225]** The term "sensor break-in" as used herein is a broad term, and is to be given its ordinary and customary meaning to a person of ordinary skill in the art (and is not to be limited to a special or customized meaning), and refers without limitation to the time (after implantation) during which the sensor's signal is becoming substantially representative of the analyte (*e.g.*, glucose) concentration (*e.g.*, where the current output from the sensor is stable relative to the glucose level). The signal may not be 'flat' when the sensor has broken-in, but in general, variation in the signal level at that point is due to a change in the analyte (*e.g.*, glucose) concentration. In some embodiments, sensor break-in occurs prior to

obtaining a meaningful calibration of the sensor output. In some embodiments, sensor break-in generally includes both electrochemical break-in and membrane break-in.

**[0226]** The term "membrane break-in" as used herein is a broad term, and is to be given its ordinary and customary meaning to a person of ordinary skill in the art (and is not to be limited to a special or customized meaning), and refers without limitation to equilibration of the membrane to its surrounding environment (*e.g.*, physiological environment *in vivo*).

**[0227]** The term "electrochemical break-in" as used herein is a broad term, and is to be given its ordinary and customary meaning to a person of ordinary skill in the art (and is not to be limited to a special or customized meaning), and refers without limitation to the time, after *in vitro* and/or *in vivo* settling of the current output from the sensor following the application of the potential to the sensor.

**[0228]** The term "host" as used herein is a broad term, and is to be given its ordinary and customary meaning to a person of ordinary skill in the art (and is not to be limited to a special or customized meaning), and refers without limitation to animals or plants, for example humans.

**[0229]** The term "continuous (or continual) analyte sensing" as used herein is a broad term, and is to be given its ordinary and customary meaning to a person of ordinary skill in the art (and is not to be limited to a special or customized meaning), and refers without limitation to the period in which monitoring of analyte concentration is continuously, continually, and or intermittently (regularly or irregularly) performed, for example, about every 5 to 10 minutes.

**[0230]** The term "electrochemically reactive surface" as used herein is a broad term, and is to be given its ordinary and customary meaning to a person of ordinary skill in the art (and is not to be limited to a special or customized meaning), and refers without limitation to a surface where an electrochemical reaction takes place. For example, a working electrode measures hydrogen peroxide produced by the enzyme-catalyzed reaction of the analyte detected, which reacts to create an electric current. Glucose analyte can be detected utilizing glucose oxidase, which produces $H_2O_2$ as a byproduct. $H_2O_2$ reacts with the surface of the working electrode, producing two protons ($2H^+$), two electrons ($2e^-$) and one molecule of oxygen ($O_2$), which produces the electronic current being detected.

**[0231]** The terms "electronic connection," "electrical connection," "electrical contact" as used herein are broad terms, and are to be given their ordinary and customary meaning to a person of ordinary skill in the art (and is not to be limited to a special or customized meaning), and refer without limitation to any connection between two electrical conductors known to those in the art. In one embodiment, electrodes are in electrical connection with the electronic circuitry of a device.

**[0232]** The term "sensing region" as used herein is a broad term, and is to be given its ordinary and customary meaning to a person of ordinary skill in the art (and is not to be limited to a special or customized meaning), and refers without limitation to the region of a monitoring device responsible for the detection of a particular analyte. The sensing region generally comprises a non-conductive body, a working electrode (anode), and can include a reference electrode (optional), and/or a counter electrode (cathode) forming electrochemically reactive surfaces on the body.

**[0233]** The term "domain" as used herein is a broad term, and is to be given its ordinary and customary meaning to a person of ordinary skill in the art (and is not to be limited to a special or customized meaning), and refers without limitation to a region of the membrane system that can be a layer, a uniform or non-uniform gradient (for example, an anisotropic region of a membrane), or a portion of a membrane.

**[0234]** The term "distal to" as used herein is a broad term, and is to be given its ordinary and customary meaning to a person of ordinary skill in the art (and is not to be limited to a special or customized meaning), and refers without limitation to the spatial relationship between various elements in comparison to a particular point of reference. In general, the term indicates an element is located relatively far from the reference point than another element.

**[0235]** The term "proximal to" as used herein is a broad term, and is to be given its ordinary and customary meaning to a person of ordinary skill in the art (and is not to be limited to a special or customized meaning), and refers without limitation to the spatial relationship between various elements in comparison to a particular point of reference. In general, the term indicates an element is located relatively near to the reference point than another element.

**[0236]** The term "*in vivo* portion" as used herein is a broad term, and is to be given its ordinary and customary meaning to a person of ordinary skill in the art (and is not to be limited to a special or customized meaning), and refers without limitation to a portion of a device (for example, a sensor) adapted for insertion into and/or existence within a living body of a host.

**[0237]** The term "*ex vivo* portion" as used herein is a broad term, and is to be given its ordinary and customary meaning to a person of ordinary skill in the art (and is not to be limited to a special or customized meaning), and refers without limitation to a portion of a device (for example, a sensor) adapted to remain and/or exist outside of a living body of a host.

**[0238]** The terms "raw data," "raw data stream", "raw data signal", "data signal", and "data stream" as used herein are broad terms, and are to be given their ordinary and customary meaning to a person of ordinary skill in the art (and are not to be limited to a special or customized meaning), and refer without limitation to an analog or digital signal from the analyte sensor directly related to the measured analyte. For example, the raw data stream is digital data in "counts" converted by an A/D converter from an analog signal (for example, voltage or amps) representative of an analyte concentration. The terms can include a plurality of time spaced data points from a substantially continuous analyte sensor, each of which comprises individual measurements taken at time intervals ranging from fractions of a second up

to, for example, 1, 2, or 5 minutes or longer. In some embodiments, the terms can refer to data that has been integrated or averaged over a time period (e.g., 5 minutes).

**[0239]** The term "count" as used herein is a broad term, and is to be given its ordinary and customary meaning to a person of ordinary skill in the art (and is not to be limited to a special or customized meaning), and refers without limitation to a unit of measurement of a digital signal. For example, a raw data stream or raw data signal measured in counts is directly related to a voltage (for example, converted by an A/D converter), which is directly related to current from the working electrode. In some embodiments, the terms can refer to data that has been integrated or averaged over a time period (e.g., 5 minutes).

**[0240]** The terms "sensor" and "sensor system" as used herein are broad terms, and are to be given their ordinary and customary meaning to a person of ordinary skill in the art (and are not to be limited to a special or customized meaning), and refer without limitation to a device, component, or region of a device by which an analyte can be quantified.

**[0241]** The term "needle" as used herein is a broad term, and is to be given its ordinary and customary meaning to a person of ordinary skill in the art (and is not to be limited to a special or customized meaning), and refers without limitation to a slender hollow instrument for introducing material into or removing material from the body.

**[0242]** The terms "operatively connected," "operatively linked," "operably connected," and "operably linked" as used herein are broad terms, and are to be given their ordinary and customary meaning to a person of ordinary skill in the art (and are not to be limited to a special or customized meaning), and refer without limitation to one or more components linked to one or more other components. The terms can refer to a mechanical connection, an electrical connection, or any connection that allows transmission of signals between the components. For example, one or more electrodes can be used to detect the amount of analyte in a sample and to convert that information into a signal; the signal can then be transmitted to a circuit. In such an example, the electrode is "operably linked" to the electronic circuitry. The terms include wired and wireless connections.

**[0243]** The terms "membrane" and "membrane system" as used herein are broad terms, and are to be given their ordinary and customary meaning to a person of ordinary skill in the art (and are not to be limited to a special or customized meaning), and refer without limitation to a permeable or semi-permeable membrane that can be comprised of one or more domains and is typically constructed of materials of one or more microns in thickness, which is permeable to oxygen and to an analyte, *e.g.*, glucose or another analyte. In one example, the membrane system comprises an immobilized glucose oxidase enzyme, which enables a reaction to occur between glucose and oxygen whereby a concentration of glucose can be measured.

**[0244]** The terms "processor module" and "microprocessor" as used herein are broad terms, and are to be given their ordinary and customary meaning to a person of ordinary skill in the art (and are not to be limited to a special or customized meaning), and refer without limitation to a computer system, state machine, processor, and the like designed to perform arithmetic or logic operations using logic circuitry that responds to and processes the basic instructions that drive a computer.

**[0245]** The term "calibration" as used herein is a broad term, and is to be given its ordinary and customary meaning to a person of ordinary skill in the art (and is not to be limited to a special or customized meaning), and refers without limitation to the relationship and/or process of determining the relationship between the sensor data and the corresponding reference data, which can be used to convert sensor data into values substantially equivalent to the reference data. In some embodiments, namely, in continuous analyte sensors, calibration can be updated or recalibrated over time if changes in the relationship between the sensor data and reference data occur, for example, due to changes in sensitivity, baseline, transport, metabolism, and the like.

**[0246]** The terms "interferants" and "interfering species" as used herein are broad terms, and are to be given their ordinary and customary meaning to a person of ordinary skill in the art (and they are not to be limited to a special or customized meaning), and refer without limitation to effects and/or species that interfere with the measurement of an analyte of interest in a sensor to produce a signal that does not accurately represent the analyte measurement. In one example of an electrochemical sensor, interfering species are compounds with an oxidation/reduction potential that overlaps with the analyte to be measured, producing a false positive signal. In another example of an electrochemical sensor, interfering species are substantially non-constant compounds (e.g., the concentration of an interfering species fluctuates over time). Interfering species include but are not limited to compounds with electroactive acidic, amine or sulfhydryl groups, urea, lactic acid, phosphates, citrates, peroxides, amino acids, amino acid precursors or break-down products, nitric oxide (NO), NO-donors, NO-precursors, acetaminophen, ascorbic acid, bilirubin, cholesterol, creatinine, dopamine, ephedrine, ibuprofen, L-dopa, methyl dopa, salicylate, tetracycline, tolazamide, tolbutamide, triglycerides, and uric acid electroactive species produced during cell metabolism and/or wound healing, electroactive species that arise during body pH changes and the like.

**[0247]** The term "single point glucose monitor" as used herein is a broad term, and is to be given its ordinary and customary meaning to a person of ordinary skill in the art (and is not to be limited to a special or customized meaning), and refers without limitation to a device that can be used to measure a glucose concentration within a host at a single point in time, for example, some embodiments utilize a small volume in vitro glucose monitor that includes an enzyme

membrane such as described with reference to U.S. Patent 4,994,167 and U.S. Patent 4,757,022. It should be understood that single point glucose monitors can measure multiple samples (for example, blood, or interstitial fluid); however only one sample is measured at a time and typically requires some user initiation and/or interaction.

**[0248]** The term "specific gravity" as used herein is a broad term, and is to be given its ordinary and customary meaning to a person of ordinary skill in the art (and is not to be limited to a special or customized meaning), and refers without limitation to the ratio of density of a material (e.g., a liquid or a solid) to the density of distilled water.

**[0249]** The terms "substantial" and "substantially" as used herein are broad terms, and are to be given their ordinary and customary meaning to a person of ordinary skill in the art (and are not to be limited to a special or customized meaning), and refer without limitation to a sufficient amount that provides a desired function. For example, an amount greater than 50 percent, an amount greater than 60 percent, an amount greater than 70 percent, an amount greater than 80 percent, or an amount greater than 90 percent.

**[0250]** The term "casting" as used herein is a broad term, and is to be given its ordinary and customary meaning to a person of ordinary skill in the art (and is not to be limited to a special or customized meaning), and refers without limitation to a process where a fluid material is applied to a surface or surfaces and allowed to cure or dry. The term is broad enough to encompass a variety of coating techniques, for example, using a draw-down machine (i.e., drawing-down), dip coating, spray coating, spin coating, and the like.

**[0251]** The term "dip coating" as used herein is a broad term, and is to be given its ordinary and customary meaning to a person of ordinary skill in the art (and is not to be limited to a special or customized meaning), and refers without limitation to coating, which involves dipping an object or material into a liquid coating substance.

**[0252]** The term "spray coating" as used herein is a broad term, and is to be given its ordinary and customary meaning to a person of ordinary skill in the art (and is not to be limited to a special or customized meaning), and refers without limitation to coating, which involves spraying a liquid coating substance onto an object or material.

**[0253]** The term "spin coating" as used herein is a broad term, and is to be given its ordinary and customary meaning to a person of ordinary skill in the art (and is not to be limited to a special or customized meaning), and refers without limitation to a coating process in which a thin film is created by dropping a raw material solution onto a substrate while it is rotating.

**[0254]** The terms "solvent" and "solvent system" as used herein are broad terms, and are to be given their ordinary and customary meaning to a person of ordinary skill in the art (and are not to be limited to a special or customized meaning), and refer without limitation to substances (e.g., liquids) capable of dissolving or dispersing one or more other substances. Solvents and solvent systems can include compounds and/or solutions that include components in addition to the solvent itself.

**[0255]** The term "baseline," "noise" and "background signal" as used herein are broad terms, and are to be given their ordinary and customary meaning to a person of ordinary skill in the art (and is not to be limited to a special or customized meaning), and refers without limitation to a component of an analyte sensor signal that is not related to the analyte concentration. In one example of a glucose sensor, the baseline is composed substantially of signal contribution due to factors other than glucose (for example, interfering species, non-reaction-related hydrogen peroxide, or other electroactive species with an oxidation/reduction potential that overlaps with hydrogen peroxide). In some embodiments wherein a calibration is defined by solving for the equation $y=mx+b$, the value of $b$ represents the baseline, or background, of the signal.

**[0256]** The terms "sensitivity" and "slope" as used herein are broad terms, and are to be given their ordinary and customary meaning to a person of ordinary skill in the art (and are not to be limited to a special or customized meaning), and refer without limitation to an amount of electrical current produced by a predetermined amount (unit) of the measured analyte. For example, in one preferred embodiment, a glucose sensor has a sensitivity (or slope) of from about 1 to about 25 picoAmps of current for every 1 mg/dL of glucose.

**[0257]** The terms "baseline and/or sensitivity shift," "baseline and/or sensitivity drift," "shift," and "drift" as used herein are broad terms, and are to be given their ordinary and customary meaning to a person of ordinary skill in the art (and are not to be limited to a special or customized meaning), and refer without limitation to a change in the baseline and/or sensitivity of the sensor signal over time. While the term "shift" generally refers to a substantially distinct change over a relatively short time period, and the term "drift" generally refers to a substantially gradual change over a relatively longer time period, the terms can be used interchangeably and can also be generally referred to as "change" in baseline and/or sensitivity.

**[0258]** The term "hypoglycemia" as used herein is a broad term, and is to be given its ordinary and customary meaning to a person of ordinary skill in the art (and are not to be limited to a special or customized meaning), and refers without limitation to a condition in which a limited or low amount of glucose exists in a host. Hypoglycemia can produce a variety of symptoms and effects but the principal problems arise from an inadequate supply of glucose as fuel to the brain, resulting in impairment of function (neuroglycopenia). Derangements of function can range from vaguely "feeling bad" to coma, and (rarely) permanent brain damage or death.

**[0259]** The term "hyperglycemia" as used herein is a broad term, and is to be given its ordinary and customary meaning

to a person of ordinary skill in the art (and are not to be limited to a special or customized meaning), and refers without limitation to a condition in which an excessive or high amount of glucose exists in a host. Hyperglycemia is one of the classic symptoms of diabetes mellitus. Non-diabetic hyperglycemia is associated with obesity and certain eating disorders, such as bulimia nervosa. Hyperglycemia is also associated with other diseases (or medications) affecting pancreatic function, such as pancreatic cancer. Hyperglycemia is also associated with poor medical outcomes in a variety of clinical settings, such as intensive or critical care settings.

**[0260]** The term "potentiostat" as used herein is a broad term, and is to be given its ordinary and customary meaning to a person of ordinary skill in the art (and is not to be limited to a special or customized meaning), and refers without limitation to an electronic instrument that controls the electrical potential between the working and reference electrodes at one or more preset values. Typically, a potentiostat works to keep the potential constant by noticing changes in the resistance of the system and compensating inversely with a change in the current. As a result, a change to a higher resistance would cause the current to decrease to keep the voltage constant in the system. In some embodiments, a potentiostat forces whatever current is necessary to flow between the working and counter electrodes to keep the desired potential, as long as the needed cell voltage and current do not exceed the compliance limits of the potentiostat.

**[0261]** The terms "electronics" and "sensor electronics" as used herein are broad terms, and are to be given their ordinary and customary meaning to a person of ordinary skill in the art (and are not to be limited to a special or customized meaning), and refer without limitation to electronics operatively coupled to the sensor and configured to measure, process, receive, and/or transmit data associated with a sensor. In some embodiments, the electronics include at least a potentiostat that provides a bias to the electrodes and measures a current to provide the raw data signal. The electronics are configured to calculate at least one analyte sensor data point. For example, the electronics can include a potentiostat, A/D converter, RAM, ROM, and/or transmitter. In some embodiments, the potentiostat converts the raw data (e.g., raw counts) collected from the sensor and converts it to a value familiar to the host and/or medical personnel. For example, the raw counts from a glucose sensor can be converted to milligrams of glucose per deciliter of blood (e.g., mg/dl). In some embodiments, the sensor electronics include a transmitter that transmits the signals from the potentiostat to a receiver (e.g., a remote analyzer, such as but not limited to a remote analyzer unit), where additional data analysis and glucose concentration determination can occur.

**[0262]** The terms "coupling" and "operatively coupling" as used herein are broad terms, and are to be given their ordinary and customary meanings to a person of ordinary skill in the art (and are not to be limited to a special or customized meaning), and refer without limitation to a joining or linking together of two or more things, such as two parts of a device or two devices, such that the things can function together. In one example, two containers can be operatively coupled by tubing, such that fluid can flow from one container to another. Coupling does not imply a physical connection. For example, a transmitter and a receiver can be operatively coupled by radio frequency (RF) transmission/communication.

**[0263]** The term "fluid communication" as used herein is a broad term, and is to be given its ordinary and customary meaning to a person of ordinary skill in the art (and are not to be limited to a special or customized meaning), and refers without limitation to two or more components (e.g., things such as parts of a body or parts of a device) functionally linked such that fluid can move from one component to another. These terms do not imply directionality.

**[0264]** The terms "continuous" and "continuously" as used herein are broad terms, and are to be given their ordinary and customary meanings to a person of ordinary skill in the art (and are not to be limited to a special or customized meaning), and refer without limitation to the condition of being marked by substantially uninterrupted extension in space, time or sequence. In one embodiment, an analyte concentration is measured continuously or continually, for example at time intervals ranging from fractions of a second up to, for example, 1, 2, or 5 minutes, or longer. It should be understood that continuous glucose sensors generally continually measure glucose concentration without required user initiation and/or interaction for each measurement, such as described with reference to U.S. Pat. N. 6,001,067, for example. These terms include situations wherein data gaps can exist (e.g., when a continuous glucose sensor is temporarily not providing data).

**[0265]** The term "medical device" as used herein is a broad term, and is to be given its ordinary and customary meaning to a person of ordinary skill in the art (and are not to be limited to a special or customized meaning), and refers without limitation to an instrument, apparatus, implement, machine, contrivance, implant, *in vitro* reagent, or other similar or related article, including a component part, or accessory which is intended for use in the diagnosis of disease or other conditions, or in the cure, mitigation, treatment, or prevention of disease, in man or other animals, or intended to affect the structure or any function of the body of man or other animals. Medical devices that can be used in conjunction with various embodiments of the analyte sensor system include any monitoring device requiring placement in a human vessel, duct or body cavity, a dialysis machine, a heart-lung bypass machine, blood collection equipment, a blood pressure monitor, an automated blood chemistry analysis device and the like.

**[0266]** The term "blood pressure monitor" as used herein is a broad term, and is to be given its ordinary and customary meaning to a person of ordinary skill in the art (and are not to be limited to a special or customized meaning), and refers without limitation to an instrument for monitoring the blood pressure of a human or other animal. For example, a blood pressure monitor can be an invasive blood pressure monitor, which periodically monitors the host's blood pressure *via*

a peripheral artery, using a blood pressure transducer, such as but not limited to a disposable blood pressure transducer. Utah Medical Products Inc. (Midvale, Utah, USA) produces a variety of Deltran® Brand disposable blood pressure transducers that are suitable for use with various embodiments disclosed herein.

**[0267]** The term "pressure transducer" as used herein is a broad term, and is to be given its ordinary and customary meaning to a person of ordinary skill in the art (and are not to be limited to a special or customized meaning), and refers without limitation to a component of an intra-arterial blood pressure monitor that measures the host's blood pressure.

**[0268]** The term "blood chemistry analysis device" as used herein is a broad term, and is to be given its ordinary and customary meaning to a person of ordinary skill in the art (and are not to be limited to a special or customized meaning), and refers without limitation to a device that measures a variety of blood components, characteristics or analytes therein. In one embodiment, a blood chemistry analysis device periodically withdraws an aliquot of blood from the host, measures glucose, $O_2$, $CO_2$, $PCO_2$, $PO_2$, potassium, sodium, pH, lactate, urea, bilirubin, creatinine, hematocrit, various minerals, and/or various metabolites, and the like, and returns the blood to the host's circulatory system. A variety of devices exist for testing various blood properties/analytes at the bedside, such as but not limited to the blood gas and chemistry devices manufactured by Via Medical (Austin, Texas, USA).

**[0269]** The term "vascular access device" as used herein is a broad term, and is to be given its ordinary and customary meaning to a person of ordinary skill in the art (and are not to be limited to a special or customized meaning), and refers without limitation to any device that is in communication with the vascular system of a host. Vascular access devices include but are not limited to catheters, shunts, blood withdrawal devices, connectors, valves, tubing and the like.

**[0270]** The term "catheter" as used herein is a broad term, and is to be given its ordinary and customary meaning to a person of ordinary skill in the art (and are not to be limited to a special or customized meaning), and refers without limitation to a tube that can be inserted into a host's body (e.g., cavity, duct or vessel). In some circumstances, catheters allow drainage or injection of fluids or access by medical instruments or devices. In some embodiments, a catheter is a thin, flexible tube (e.g., a "soft" catheter). In alternative embodiments, the catheter can be a larger, solid tube (e.g., a "hard" catheter). The term "cannula" is interchangeable with the term "catheter" herein.

**[0271]** The term "indwell" as used herein is a broad term, and is to be given its ordinary and customary meaning to a person of ordinary skill in the art (and are not to be limited to a special or customized meaning), and refers without limitation to reside within a host's body. Some medical devices can indwell within a host's body for various lengths of time, depending upon the purpose of the medical device, such as but not limited to a few hours, days, weeks, to months, years, or even the host's entire lifetime. In one exemplary embodiment, an arterial catheter may indwell within the host's artery for a few hours, days, a week, or longer, such as but not limited to the host's perioperative period (e.g., from the time the host is admitted to the hospital to the time he is discharged).

**[0272]** The term "sheath" as used herein is a broad term, and is to be given its ordinary and customary meaning to a person of ordinary skill in the art (and are not to be limited to a special or customized meaning), and refers without limitation to a covering or supporting structure that fits closely around something, for example, in the way that a sheath covers a blade. In one exemplary embodiment, a sheath is a slender, flexible, polymer tube that covers and supports a wire-type sensor prior to and during insertion of the sensor into a catheter.

**[0273]** The term "slot" as used herein is a broad term, and is to be given its ordinary and customary meaning to a person of ordinary skill in the art (and are not to be limited to a special or customized meaning), and refers without limitation to a relatively narrow opening.

**[0274]** The term "regulator" as used herein is a broad term, and is to be given its ordinary and customary meaning to a person of ordinary skill in the art (and are not to be limited to a special or customized meaning), and refers without limitation to a device that regulates the flow of a fluid or gas. For example, a regulator can be a valve or a pump.

**[0275]** The term "pump" as used herein is a broad term, and is to be given its ordinary and customary meaning to a person of ordinary skill in the art (and are not to be limited to a special or customized meaning), and refers without limitation to a device used to move liquids, or slurries. In general, a pump moves liquids from lower pressure to higher pressure, and overcomes this difference in pressure by adding energy to the system (such as a water system).

**[0276]** The term "valve" as used herein is a broad term, and is to be given its ordinary and customary meaning to a person of ordinary skill in the art (and are not to be limited to a special or customized meaning), and refers without limitation to a device that regulates the flow of substances (either gases, fluidized solids, slurries, or liquids), for example, by opening, closing, or partially obstructing a passageway through which the substance flows. In general, a valve allows no flow, free flow and/or metered flow through movement of the valve between one or more discreet positions.

**[0277]** The term "retrograde" as used herein is a broad term, and is to be given its ordinary and customary meaning to a person of ordinary skill in the art (and are not to be limited to a special or customized meaning), and refers without limitation to orientation (e.g., of a catheter) against the direction of blood flow.

**[0278]** The term "antegrade" as used herein is a broad term, and is to be given its ordinary and customary meaning to a person of ordinary skill in the art (and are not to be limited to a special or customized meaning), and refers without limitation to orientation (e.g., of a catheter) with the direction of blood flow.

**[0279]** The term "biological sample" as used herein is a broad term, and is to be given its ordinary and customary

meaning to a person of ordinary skill in the art (and are not to be limited to a special or customized meaning), and refers without limitation to any biological material to be tested for the presence and/or concentration of an analyte in a sample. Examples biological samples that may be tested include blood, serum, plasma, saliva, urine, ocular fluid, semen, and spinal fluid, tissue, and the like.

**[0280]** The term "diffusion barrier" as used herein is a broad term, and is to be given its ordinary and customary meaning to a person of ordinary skill in the art (and is not to be limited to a special or customized meaning) and refers without limitation to something that obstructs the random movement of compounds, species, atoms, molecules, or ions from one site in a medium to another. In some embodiments, a diffusion barrier is structural, such as a wall that separates two working electrodes and substantially prevents diffusion of a species from one electrode to the other. In some embodiments, a diffusion barrier is spatial, such as separating working electrodes by a distance sufficiently large enough to substantially prevent a species at a first electrode from affecting a second electrode. In other embodiments, a diffusion barrier can be temporal, such as by turning the first and second working electrodes on and off, such that a reaction at a first electrode will not substantially affect the function of the second electrode.

**[0281]** The term "constant analyte" as used herein is a broad term, and is to be given its ordinary and customary meaning to a person of ordinary skill in the art (and it is not to be limited to a special or customized meaning), and refers without limitation to an analyte that remains relatively constant over a time period, for example over an hour to a day as compared to other variable analytes. For example, in a person with diabetes, oxygen and urea may be relatively constant analytes in particular tissue compartments relative to glucose, which is known to oscillate between about 40 and 400 mg/dL during a 24-hour cycle. Although analytes such as oxygen and urea are known to oscillate to a lesser degree, for example due to physiological processes in a host, they are substantially constant, relative to glucose, and can be digitally filtered, for example low pass filtered, to minimize or eliminate any relatively low amplitude oscillations. Constant analytes other than oxygen and urea are also contemplated.

**[0282]** The terms "insulative properties," "electrical insulator" and "insulator" as used herein are broad terms, and are to be given their ordinary and customary meaning to a person of ordinary skill in the art (and is not to be limited to a special or customized meaning) and refers without limitation to the tendency of materials that lack mobile charges to prevent movement of electrical charges between two points. In one exemplary embodiment, an electrically insulative material may be placed between two electrically conductive materials, to prevent movement of electricity between the two electrically conductive materials. In some embodiments, the terms refer to a sufficient amount of insulative property (e.g., of a material) to provide a necessary function (electrical insulation). The terms "insulator" and "non-conductive material" can be used interchangeably herein.

**[0283]** The terms "inactive enzyme" or "inactivated enzyme" as used herein are broad terms, and are to be given their ordinary and customary meaning to a person of ordinary skill in the art (and it is not to be limited to a special or customized meaning), and refer without limitation to an enzyme (e.g., glucose oxidase, GOx) that has been rendered inactive (e.g., "killed" or "dead") and has no enzymatic activity. Enzymes can be inactivated using a variety of techniques known in the art, such as but not limited to heating, freeze-thaw, denaturing in organic solvent, acids or bases, cross-linking, genetically changing enzymatically critical amino acids, and the like. In some embodiments, a solution containing active enzyme can be applied to the sensor, and the applied enzyme subsequently inactivated by heating or treatment with an inactivating solvent.

**[0284]** The term "non-enzymatic" as used herein is a broad term, and is to be given their ordinary and customary meaning to a person of ordinary skill in the art (and it is not to be limited to a special or customized meaning), and refers without limitation to a lack of enzyme activity. In some embodiments, a "non-enzymatic" membrane portion contains no enzyme; while in other embodiments, the "non-enzymatic" membrane portion contains inactive enzyme. In some embodiments, an enzyme solution containing inactive enzyme or no enzyme is applied.

**[0285]** The term "coaxial" as used herein is a broad term, and is to be given its ordinary and customary meaning to a person of ordinary skill in the art (and it is not to be limited to a special or customized meaning), and refers without limitation to having a common axis, having coincident axes or mounted on concentric shafts.

**[0286]** The term "twisted" as used herein is a broad term, and is to be given its ordinary and customary meaning to a person of ordinary skill in the art (and it is not to be limited to a special or customized meaning), and refers without limitation to united by having one part or end turned in the opposite direction to the other, such as, but not limited to the twisted strands of fiber in a string, yarn, or cable.

**[0287]** The term "helix" as used herein is a broad term, and is to be given its ordinary and customary meaning to a person of ordinary skill in the art (and it is not to be limited to a special or customized meaning), and refers without limitation to a spiral or coil, or something in the form of a spiral or coil (e.g. a corkscrew or a coiled spring). In one example, a helix is a mathematical curve that lies on a cylinder or cone and makes a constant angle with the straight lines lying in the cylinder or cone. A "double helix" is a pair of parallel helices intertwined about a common axis, such as but not limited to that in the structure of DNA.

**[0288]** The terms "integral," "integrally," "integrally formed," integrally incorporated," "unitary" and "composite" as used herein are broad terms, and are to be given their ordinary and customary meaning to a person of ordinary skill in the art

(and they are not to be limited to a special or customized meaning), and refer without limitation to the condition of being composed of essential parts or elements that together make a whole. The parts are essential for completeness of the whole. In one exemplary embodiment, at least a portion (e.g., the *in vivo* portion) of the sensor is formed from at least one platinum wire at least partially covered with an insulative coating, which is at least partially helically wound with at least one additional wire, the exposed electroactive portions of which are covered by a membrane system (see description of Fig. 1B or 9B); in this exemplary embodiment, each element of the sensor is formed as an integral part of the sensor (e.g., both functionally and structurally).

**[0289]** The term "constant noise" and "constant background" as used herein are broad terms, and are to be given their ordinary and customary meaning to a person of ordinary skill in the art (and it is not to be limited to a special or customized meaning), and refer without limitation to the component of the background signal that remains relatively constant over time. For example, certain electroactive compounds found in the human body are relatively constant factors (e.g., baseline of the host's physiology) and do not significantly adversely affect accuracy of the calibration of the glucose concentration (e.g., they can be relatively constantly eliminated using the equation $y=mx+b$). In some circumstances, constant background noise can slowly drift over time (e.g., increases or decreases), however this drift need not adversely affect the accuracy of a sensor, for example, because a sensor can be calibrated and re-calibrated and/or the drift measured and compensated for.

**[0290]** The term "non-constant noise" or non-constant background" as used herein are broad terms, and are to be given their ordinary and customary meaning to a person of ordinary skill in the art (and it is not to be limited to a special or customized meaning), and refer without limitation to a component of the background signal that is relatively non-constant, for example, transient and/or intermittent. For example, certain electroactive compounds, are relatively non-constant (e.g., intermittent interferents due to the host's ingestion, metabolism, wound healing, and other mechanical, chemical and/or biochemical factors), which create intermittent (e.g., non-constant) "noise" on the sensor signal that can be difficult to "calibrate out" using a standard calibration equations (e.g., because the background of the signal does not remain constant).

**[0291]** The terms "small diameter sensor," "small structured sensor," and "micro-sensor" as used herein are broad terms, and are to be given their ordinary and customary meaning to a person of ordinary skill in the art (and are not to be limited to a special or customized meaning), and refer without limitation to sensing mechanisms that are less than about 2 mm in at least one dimension, and more preferably less than about 1 mm in at least one dimension. In some embodiments, the sensing mechanism (sensor) is less than about 0.95, 0.9, 0.85, 0.8, 0.75, 0.7, 0.65, 0.6, 0.5, 0.4, 0.3, 0.2, or 0.1 mm. In some embodiments, the sensing mechanism is a needle-type sensor, wherein the diameter is less than about 1 mm (see, for example, U.S. Patent No. 6,613,379 to Ward et al. and in U.S. Patent Publication No. US-2006-0020187-A1). In some alternative embodiments, the sensing mechanism includes electrodes deposited on a planar substrate, wherein the thickness of the implantable portion is less than about 1 mm, see, for example U.S. Patent No. 6,175,752 to Say et al. and U.S. Patent No. 5,779,665 to Mastrototaro et al.

**[0292]** The term "GOx" as used herein is a broad term, and is to be given their ordinary and customary meaning to a person of ordinary skill in the art (and it is not to be limited to a special or customized meaning), and refers without limitation to the enzyme Glucose Oxidase (e.g., GOx is an abbreviation).

Overview

**[0293]** Intensive care medicine or critical care medicine is concerned with providing greater than ordinary medical care and/or observation to people in a critical or unstable condition. In recent years, an increasingly urgent need has arisen, for more intensive care medicine. People requiring intensive care include those recovering after major surgery, with severe head trauma, life-threatening acute illness, respiratory insufficiency, coma, haemodynamic insufficiency, severe fluid imbalance or with the failure of one or more of the major organ systems (life-critical systems or others). More than 5 million people are admitted annually to intensive care units (ICUs) and critical care units (CCUs) in the United States.

**[0294]** Intensive care is generally the most expensive, high technology and resource intensive area of medical care. In the United States estimates of the year 2000 expenditure for critical care medicine ranged from $15-55 billion accounting for about 0.5% of GDP and about 13% of national health care expenditure. As the U.S. population ages, these costs will increase substantially. Accordingly, there is an urgent need to reducing costs while at the same time reducing ICU/CCU mortality rates by improving care. Some embodiments disclosed herein are suitable for use in an intensive care or critical care unit of a medical care facility for substantially continuously measuring a host's analyte concentration.

**[0295]** Hyperglycemia is a medical condition in which an excessive amount of glucose circulates in a host. Medical studies suggest a relationship between hyperglycemia and host outcome in intensive/critical care settings. For example, perioperative hyperglycemia is associated with increased rates and severity of myocardial infarction (MI) and stroke, while tight glucose control with intravenous (IV) insulin therapy is linked to a 30% reduction in mortality one year after admission for acute MI. Furthermore, strict in-hospital glucose control is associated with 40% reductions of morbidity, mortality, sepsis, dialysis, blood transfusions, as well as reduced length of stay, reduced costs and the like.

**[0296]** Hyperglycemia can also be an issue in non-critical care settings, such as in the general hospital population, such as for diabetes hosts admitted for non-glucose-related medical conditions, or in clinical settings, such as the doctor's office, such as during glucose challenge tests, or treatment of the elderly or the very young, or others who may have difficulty with glucose control.

**[0297]** Unfortunately, using generally available technology, tight glucose control requires frequent monitoring of the host by the clinical staff, IV insulin or injections, and on-time feeding. Frequent monitoring typically requires a nurse or other staff member to measure the host's glucose concentration using a lancet (to obtain a blood sample) and a hand held glucose monitor. The nurse can perform this task many times a day (e.g., every hour or more frequently). This task becomes an undue burden that takes the nurse away from his/her other duties, or requires extra staff. The preferred embodiments disclose systems and methods to reduce and/or minimize the interaction required to regularly (e.g., continuously) measure the host's glucose concentration.

**[0298]** Unfortunately it has been shown that an effort to maintain tight control of glucose levels (e.g., about 80-129 mg/dl) can increase the risk of hypoglycemia using conventional systems and methods. For example, administration of insulin, quality, and timing of meal ingestion, and the like can lead to hypoglycemia. Because hypoglycemia can cause shock and death (immediate problems), the clinical staff rigorously avoids it, often by maintaining the host at elevated blood glucose concentrations (which can degrade the clinical outcome in the long run) and causes the problems of hyperglycemia discussed above.

**[0299]** Accordingly, in spite of clinically demonstrated improvements associated with tight glucose control, institutions are slow to adopt the therapy due to the increased workload on the staff as well as a pervasive fear of hypoglycemia, which is potentially life ending. Therefore, there is an urgent need for devices and methods that offer continuous, robust glucose monitoring, to improve patient care and lower medical costs. The preferred embodiments describe systems and methods for providing continuous glucose monitoring while providing alarms or alerts that aid in avoiding hypoglycemic events.

**[0300]** Hyperglycemia can be managed in a variety of ways. Currently, for hosts in an intensive care setting, such as and ICU, CCU or emergency room (ER), hyperglycemia is managed with sliding-scale IV insulin that stops insulin delivery at about 150 to 200 mg/dl. This generally requires monitoring by a nurse (using a hand-held clinical glucose meter) and insulin administration at least every six hours. Maintaining tight glucose control within the normal range (e.g., 80-110 mg/dl) currently requires hourly or even more frequent monitoring and insulin administration. This places an undue burden on the nursing staff. The preferred embodiments provide devices and methods for automated, continuous glucose monitoring (e.g., indwelling in the circulatory system), to enable tight glucose control.

**[0301]** The *in vivo* continuous analyte monitoring system of the preferred embodiments can be used in clinical settings, such as in the hospital, the doctor's office, long-term nursing facilities, or even in the home. The present device can be used in any setting in which frequent or continuous analyte monitoring is desirable. For example, in the ICU, hosts are often recovering from serious illness, disease, or surgery, and control of host glucose levels is important for host recovery. Use of a continuous glucose sensor as described in the preferred embodiments allows tight control of host glucose concentration and improved host care, while reducing hypoglycemic episodes and reducing the ICU staff work load. For example, the system can be used for the entire hospital stay or for only a part of the hospital stay.

**[0302]** In another example, the continuous glucose monitor of the preferred embodiments can be used in an ER setting. In the ER, a host may be unable to communicate with the staff. Routine use of a continuous analyte monitor (e.g., glucose, creatinine, phosphate, electrolytes, or drugs) can enable the ER staff to monitor and respond to analyte concentration changes indicative of the host's condition (e.g., the host's glucose concentration) without host input.

**[0303]** In yet another example, a continuous analyte monitor can be used in the general hospital population to monitor host analyte concentrations, for various lengths of time, such as during the entire hospital stay or for a portion of the hospital stay (e.g., only during surgery). For example, a diabetic host's glucose concentration can be monitored during his entire stay. In another example, a cardiac host's glucose can be monitored during surgery and while in the ICU, but not after being moved to the general host population. In another example, a jaundiced newborn infant can have his bilirubin concentration continuously monitored by an in-dwelling continuous analyte monitor until the condition has receded.

**[0304]** In addition to use in the circulatory system, the analyte sensor of the preferred embodiments can be used in other body locations. In some embodiments, the sensor is used subcutaneously. In another embodiment, the sensor can be used intracranially. In another embodiment, the sensor can be used within the spinal compartment, such as but not limited to the epidural space. In some embodiments, the sensor of the preferred embodiments can be used with or without a catheter.

Applications/Uses

**[0305]** One aspect of the preferred embodiments provides a system for *in vivo* continuous analyte monitoring (e.g., glucose, $O_2$, $CO_2$, $PCO_2$, $PO_2$, potassium, sodium, pH, lactate, urea, bilirubin, creatinine, hematocrit, various minerals,

various metabolites, and the like) that can be operatively coupled to a catheter to measure analyte concentration within the host's blood stream. In some embodiments, the system includes an analyte sensor that extends a short distance into the blood stream (e.g., out of the catheter) without substantially occluding the catheter or the host's blood stream. The catheter can be fluidly coupled to additional IV and diagnostic devices, such as a saline bag, an automated blood pressure monitor, or a blood chemistry monitor device. In some embodiments, blood samples can be removed from the host *via* the sensor system, as described elsewhere herein. In one embodiment, the sensor is a glucose sensor, and the medical staff monitors the host's glucose level.

**[0306]** **Figs. 1A** to **1E** illustrate one embodiment of an exemplary analyte sensor system **10** for measuring an analyte (e.g., glucose, urea, potassium, pH, proteins, etc.) that includes a catheter **12** configured to be inserted or pre-inserted into a host's blood stream. In clinical settings, catheters are often inserted into hosts to allow direct access to the circulatory system without frequent needle insertion (e.g., venipuncture). Suitable catheters can be sized as is known and appreciated by one skilled in the art, such as but not limited to from about French (0.33mm) or less to about 30 French (10mm) or more; and can be, for example, 2, 3, 4, 5, 6, 7, 8, 9, 10, 11, 12, 13, 14, 15, 16, 17, 18, 19, or 20 French (3 French is equivalent to about 1 mm) and/or from about 33 gauge or less to about 16 gauge or more, for example, 33, 32, 31, 30, 29, 28, 27, 26, 25, 24, 23, 22, 21, 20, 19, 18, 17, or 16 gauge. Additionally, the catheter can be shorter or longer, for example 0.75, 1.0, 1.25, 1.5, 1.75, 2.0 inches in length or longer. In some embodiments, the catheter is a venous catheter. In other embodiments, the catheter is configured for insertion into a peripheral or a central artery. In some embodiments, the catheter is configured to extend from a peripheral artery to a central portion of the host's circulatory system, such as but not limited to the heart. The catheter can be manufactured of any medical grade material known in the art, such as but not limited to polymers and glass as described herein. A catheter can include a single lumen or multiple lumens. A catheter can include one or more perforations, to allow the passage of host fluid through the lumen of the catheter.

**[0307]** The terms "inserted" or "pre-inserted" as used herein are broad terms, and are to be given their ordinary and customary meaning to a person of ordinary skill in the art (and are not to be limited to a special or customized meaning), and refer without limitation to insertion of one thing into another thing. For example, a catheter can be inserted into a host's blood stream. In some embodiments, a catheter is "pre-inserted," meaning inserted before another action is taken (e.g., insertion of a catheter into a host's blood stream prior to insertion of a sensor into the catheter). In some exemplary embodiments, a sensor is coupled to a pre-inserted catheter, namely, one that has been previously inserted (or pre-inserted) into the host's circulatory system.

**[0308]** Referring now to **Figs. 1A** to **1E,** in some embodiments, the catheter **12** is a thin, flexible tube having a lumen **12a,** such as is known in the art. In some embodiments, the catheter can be rigid; in other embodiments, the catheter can be custom manufactured to desired specifications (e.g., rigidity, dimensions, etc). The catheter can be a single-lumen catheter or a multi-lumen catheter. At the catheter's proximal end is a small orifice **12b** for fluid connection of the catheter to the blood stream. At the catheter's distal end is a connector **18,** such as a leur connector or other fluid connector known in the art.

**[0309]** The illustrations of **Figs. 1A** to **1E** show one exemplary embodiment of the connector **18** including a flange **18a** and a duct **18b.** In the exemplary embodiment, the flange **18a** is configured to enable connection of the catheter to other medical equipment (e.g., saline bag, pressure transducer, blood chemistry device, and the like) or capping (e.g., with a bung and the like). Although one exemplary connector is shown, one skilled in the art appreciates a variety of standard or custom made connectors suitable for use with the preferred embodiments. The duct **18b** is in fluid communication with the catheter lumen and terminates in a connector orifice **18c.**

**[0310]** In some embodiments, the catheter is inserted into the host's blood stream, such as into a vein or artery by any useful method known in the art. Generally, prior to and during insertion, the catheter is supported by a hollow needle or trochar (not shown). For example, the supported catheter can be inserted into a peripheral vein or artery, such as in the host's arm, leg, hand, or foot. Typically, the supporting needle is removed (e.g., pulled out of the connector) and the catheter is connected (e.g., *via* the connector **18**) to IV tubing and a saline drip, for example. However, in one embodiment, the catheter is configured to operatively couple to medical equipment, such as but not limited to a sensor system of the preferred embodiments. Additionally and/or alternatively, the catheter can be configured to operatively couple to another medical device, such as a pressure transducer, for measurement of the host's blood pressure.

**[0311]** In some embodiments, the catheter and the analyte sensor are configured to indwell within the host's blood stream *in vivo.* An indwelling medical device, such as a catheter or implant, is disposed within a portion of the body for a period of time, from a few minutes or hours to a few days, months, or even years. An indwelling catheter is typically inserted within a host's vein or artery for a period of time, often 2 or more days, a month, or even a few months. In some embodiments, the catheter can indwell in a host's artery or vein for the length of a perioperative period (e.g., the entire hospital stay) or for shorter or longer periods. In some embodiments, the use of an indwelling catheter permits continuous access of an analyte sensor to a blood stream while simultaneously allowing continuous access to the host's blood stream for other purposes, for example, the administration of therapeutics (e.g., fluids, drugs, etc.), measurement of physiologic properties (e.g., blood pressure), fluid removal, and the like.

**[0312]** Referring again to **Figs. 1A** to **1E,** the system **10** also includes an analyte sensor **14** configured to extend through the catheter lumen **12a** (see **Fig. 1E**), out of the catheter orifice **12b** and into the host's blood stream by about 0.010 inches to about 1 inch, or shorter or longer lengths. In some embodiments, however, the sensor may not extend out of the catheter, for example, can reside just inside the catheter tip. The sensor can extend through the catheter in any functional manner. In some embodiments, the sensor is configured to be held (e.g., disposed) on an inner surface (e.g., the lumen) or outer surface of the catheter. In some embodiments, the sensor is deposited (e.g., formed) on a surface of the catheter. In some embodiments, a sensor is attached to a surface of the catheter, such as by an adhesive and/or welding. In some other embodiments, the sensor is configured to "free float" within the lumen of the catheter.

**[0313]** In some embodiments, the sensor **14** is configured to measure the concentration of an analyte (e.g., glucose, $O_2$, $CO_2$, $PCO_2$, $PO_2$, potassium, sodium, pH, lactate, urea, bilirubin, creatinine, hematocrit, various minerals, various metabolites, and the like) within the host's blood stream. In some preferred embodiments, the sensor includes at least one electrode (see, e.g., **Fig. 3B**), for example a working electrode; however any combination of working electrode(s), reference electrode(s), and/or counter electrode(s) can be implemented as is appreciated by one skilled in the art. For example, in some preferred embodiments, the sensor includes at least two working electrodes, as is described with reference to **Figs. 3D** through **3I.** Preferably, the sensor **14** includes at least one exposed electroactive area (e.g., working electrode), a membrane system (e.g., including an enzyme), a reference electrode (proximal to or remote from the working electrode), and an insulator material. Various systems and methods for design and manufacture of continuous analyte sensors are described in more detail elsewhere herein. In some embodiments, the sensor is a needle-type continuous analyte sensor, configured as disclosed in U.S. Patent Publication No. US-2006-0020192-A1 and U.S. Patent Publication No. US-2006-0036143-A1, both of which are incorporated herein by reference in their entirety. In some embodiments, the sensor is configured to measure glucose concentration. Exemplary sensor configurations are discussed in more detail, elsewhere herein.

**[0314]** Referring to **Figs 1A** to **1E,** the sensor has a proximal end **14a** and a distal end **14b.** At its distal end **14b,** the sensor **14** is associated with (e.g., connected to, held by, extends through, and the like) a fluid coupler **20** having first and second sides (**20a** and **20b,** respectively). The fluid coupler is configured to mate (*via* its first side **20a**) to the catheter connector **18.** In one embodiment, a skirt **20c** is located at the fluid coupler's first side and includes an interior surface **20d** with threads **20e** (see **Figs. 1D** and **1E**). In this embodiment, the fluid coupler is configured to mate with the connector flange **18a,** which is screwed into the fluid coupler *via* the screw threads. However, in other embodiments, the fluid coupler is configured to mate with the connector using any known mating configuration, for example, a snap-fit, a press-fit, an interference-fit, and the like, and can include a locking mechanism to prevent separation of the connector and fluid coupler. The fluid coupler **20** includes a lumen **20f** extending from a first orifice **20h** on its first side **20a** to a second orifice **20i** located on the fluid coupler's second side **20b** (**Figs. 1C1** to **1E**). When the catheter connector is mated with the fluid coupler, the catheter's lumen **12a** is in fluid communication with the fluid coupler's lumen **20f** *via* orifices **18c** and **20h.**

**[0315]** **Figs. 1A** to **1D** show one embodiment of a fluid coupler **20,** namely, a Y-coupler; however, any known coupler configuration can be used, including but not limited to a straight coupler, a T-coupler, a cross-coupler, a custom configured coupler, and the like. In some embodiments, the fluid coupler includes at least one valve (e.g., a septum, a 3-way valve, a stop-cock valve), which can be used for a variety of purposes (e.g., injection of drugs). The fluid coupler can be made of any convenient material, such as but not limited to plastic, glass, metal or combinations thereof and can be configured to withstand known sterilization techniques.

**[0316]** In the exemplary embodiment, the second side **20b** of the fluid coupler **20** is configured to be operably connected to IV equipment, another medical device or to be capped, and can use any known mating configuration, for example, a snap-fit, a press-fit, an interference-fit, and the like. In one exemplary embodiment, the second side **20b** is configured to mate with a saline drip, for delivery of saline to the host. For example, the saline flows from an elevated bag of sterile saline *via* tubing, through the fluid coupler, through the catheter and into the host's blood system (e.g., vein or artery). In another embodiment, a syringe can be mated to the fluid coupler, for example, to withdraw blood from the host, *via* the catheter. Additional connection devices (e.g., a three-way valve) can be operably connected to the fluid coupler, to support additional functionality and connection of various devices, such as but not limited to a blood pressure transducer.

**[0317]** Referring to the exemplary embodiment of **Figs. 1A** and **1E,** at least a portion of the sensor **14** passes through the fluid coupler **20** (e.g., the fluid coupler lumen **20f**) and is operatively connected to sensor electronics (not shown) *via* a hardwire **24.** In alternative embodiments however, the sensor electronics can be disposed in part or in whole with the fluid coupler (e.g., integrally with or proximal to) or can be disposed in part or in whole remotely from the fluid coupler (*e.g.*, on a stand or at the bed side). Connections between the sensor and sensor electronics (in part or in whole) can be accomplished using known wired or wireless technology. In one exemplary embodiment, the sensor is hardwired to the electronics located substantially wholly remote from the fluid coupler (e.g., disposed on a stand or near the bedside); one advantage of remote electronics includes enabling a smaller sized fluid coupler design. In another exemplary embodiment, a portion of the sensor electronics, such as a potentiostat, is disposed on the fluid coupler and the remaining electronics (e.g., electronics for receiving, data processing, printing, connection to a nurses' station, etc.) are disposed

remotely from the fluid coupler (e.g., on a stand or near the bedside). One advantage of this design can include more reliable electrical connection with the sensor in some circumstances. In this embodiment, the potentiostat can be hard-wired directly to the remaining electronics or a transmitter can be disposed on or proximal to the fluid coupler, for remotely connecting the potentiostat to the remaining electronics (e.g., by radio frequency (RF)). In another exemplary embodiment, all of the sensor electronics can be disposed on the fluid coupler. In still another embodiment, the sensor electronics disposed on the fluid coupler include a potentiostat.

**[0318]** Referring again to **Figs. 1A** to **1E,** a protective sheath **26** is configured to cover at least a portion of the sensor **14** during insertion, and includes hub **28** and slot **30.** In general, the protective sheath protects and supports the sensor prior to and during insertion into the catheter **12** *via* the connector **18.** The protective sheath can be made of biocompatible polymers known in the art, such as but not limited to polyethylene (PE), polyurethane (PE), polyvinyl chloride (PVC), polycarbonate (PC), nylon, polyamides, polyimide, polytetrafluoroethylene (PTFE), Teflon, nylon and the like. The protective sheath includes a hub **28,** for grasping the sheath (e.g., while maintaining sterilization of the sheath). In this embodiment, the hub additionally provides for mating with the second side **20b** of the fluid coupler **20,** prior to and during sensor insertion into the catheter. In this exemplary embodiment, the slot of the protective sheath is configured to facilitate release of the sensor therefrom. In this embodiment, after the sensor has been inserted into the catheter, the hub is grasped and pulled from the second side of the fluid coupler. This action peels the protective sheath from the sensor (e.g., the sensor slides through the slot as the sheath is removed), leaving the sensor within the catheter. The second side of the fluid coupler can be connected to other medical devices (e.g., a blood pressure monitor) or an IV drip (e.g., a saline drip), or capped. In alternative embodiments, the sheath can fold (e.g., fold back or concertinas) or retract (e.g., telescope) during insertion, to expose the sensor. In other embodiments, the sheath can be configured to tear away from the sensor before, during, or after insertion of the sensor. In still other embodiments, the sheath can include an outlet hole **30a,** to allow protrusion of the sensor from the back end of the sheath (e.g., near the hub **28**). One skilled in the art will recognize that additional configurations can be used, to separate the sensor **14** from the sheath **26.**

**[0319]** In some embodiments, the sensor includes at least two working electrodes **14,** which can be twisted and/or bundled, such as in a helical and/or coaxial configuration. In some embodiments, the two working electrodes are twisted into a "twisted pair," which can be configured to be inserted into and to extend within a vascular access device, such as a catheter **12** or cannula implanted in a host's vein or artery, as is described in more detail in the section entitled "Integrated Sensor System." In some embodiments, the twisted pair is configured to reside within the lumen **12a** of the catheter **12;** while in other embodiments, the twisted pair is configured to protrude from the catheter's proximal orifice **12b.** In still other embodiments, the twisted pair is configured to intermittently protrude from the catheter's proximal orifice **12b**.

**[0320]** In some embodiments, the sheath **26** can be optional, depending upon the sensor design. For example, the sensor can be inserted into a catheter or other vascular access device with or without the use of a protective sheath). In some embodiments, the sensor can be disposed on the outer surface of a catheter (as described elsewhere herein) or on the inner surface of a catheter; and no sheath is provided. In other embodiments, a multi-lumen catheter can be provided with a sensor already disposed within one of the lumens; wherein the catheter is inserted into the host's vein or artery with the sensor already disposed in one of the lumens.

**[0321]** In some alternative embodiments, an analyte sensor is integrally formed on a catheter. In various embodiments, the catheter can be placed into a host's vein or artery in the usual way a catheter is inserted, as is known by one skilled in the art, and the host's analyte concentration measured substantially continuously. In some embodiments, the sensor system can be coupled to one or more additional devices, such as a saline bag, an automated blood pressure monitor, a blood chemistry monitor device, and the like. In one exemplary embodiment, the integrally formed analyte sensor is a glucose sensor.

**[0322]** **Figs. 2A** to **2B** illustrate one exemplary embodiment of an analyte sensor integrally formed on a catheter. The system **210** is configured to measure an analyte (e.g., glucose, $O_2$, $CO_2$, $PCO_2$, $PO_2$, potassium, sodium, pH, lactate, urea, bilirubin, creatinine, hematocrit, various minerals, various metabolites, and the like) and generally includes a catheter **212** configured for insertion into a host's blood stream (e.g., *via* a vein or artery) and a sensor at least partially integrally formed on the catheter's exterior surface **232.** Preferably, the sensor **214** includes at least one exposed electroactive area **240** (e.g., a working electrode), a membrane system (e.g., including an enzyme), a reference electrode (proximal to or remote from the working electrode), and an insulator.

**[0323]** In this embodiment, the catheter includes a lumen **212a** and an orifice **212b** at its proximal end, for providing fluid connection from the catheter's lumen to the host's blood stream (see **Fig. 2A**).

**[0324]** In some embodiments, the catheter is inserted into a vein, as described elsewhere herein. In other embodiments, the catheter is inserted into an artery, as described elsewhere herein. The catheter can be any type of venous or arterial catheter commonly used in the art (e.g., peripheral catheter, central catheter, Swan-Gantz catheter, etc.). The catheter can be made of any useful medical grade material (e.g., polymers and/or glass) and can be of any size, such as but not limited to from about 1 French (0.33mm) or less to about 30 French (10mm) or more; for example, 2, 3, 4, 5, 6, 7, 8, 9, 10, 11, 12, 13, 14, 15, 16, 17, 18, 19, or 20 French (3 French is equivalent to about 1 mm). In certain embodiments, the catheter can be a single lumen catheter or a multi-lumen catheter. In some embodiments, the catheter can include one

or more perforations, to allow the passage of host fluid through the lumen of the catheter.

**[0325]** At its distal end **212c,** the catheter **212** includes (e.g., in fluid communication) a connector **218.** The connector can be of any known type, such as a leur lock, a T-connector, a Y-connector, a cross-connector or a custom configuration, for example. In some embodiments, the connector includes at least one valve. At a second side **218e** (e.g., back end), the connector **218** can be operatively connected to a saline system (e.g., saline bag and tubing), other medical devices (e.g., automatic blood chemistry machine, dialysis machine, a blood bag for collecting donated blood, etc.), or capped.

**[0326]** In some embodiments, the system **210** includes sensor electronics (not shown) operatively connected to the analyte sensor, wherein the sensor electronics are generally configured to measure and/or process the sensor data as described in more detail elsewhere herein. In some embodiments, the sensor electronics can be partially or wholly disposed with (e.g., integral with, disposed on, or proximal to) the connector **218** at the distal end of the catheter or partially or wholly remote from the catheter (e.g., on a stand or on the bedside). In one embodiment, the sensor electronics disposed with the connector include a potentiostat. In some embodiments, the sensor electronics are configured to measure the host's analyte concentration substantially continuously. For example, the sensor can measure the analyte concentration continuously or at time intervals ranging from fractions of a second up to, for example, 1, 2, or 5 minutes or longer.

**[0327]** **Figs. 2C** to **2F** illustrate additional embodiments of the sensor shown in **Figs. 2A** to **2B.** The catheter **212** is shown with an integral sensor **214** having at least one electrode **240** formed on its exterior surface **232** (e.g., **Fig. 2F**). In general, the sensor can be designed with 1, 2, 3, 4 or more electrodes and can be connected by traces (or the like) to electrical contacts **218d** (or the like) at the second end of the connector **218** (e.g., **Figs. 2A** to **2F**). In some embodiments, the sensor is hard-wired to the sensor electronics; alternatively, any operable connection can be used. Preferably, the sensor includes at least one working electrode and at least one reference or counter electrode. In some embodiments, the reference electrode is located proximal to the at least one working electrode (e.g., adjacent to or near to the working electrode). In some alternative embodiments, the reference electrode is located remotely from the working electrode (e.g., away from the working electrode, such as but not limited to within the lumen of the catheter **212** (or connector **218**), on the exterior of the sensor system, in contact with the patient (e.g., on the skin), or the like). In some embodiments, the reference electrode is located proximal to or within the fluid connector, such as but not limited to, coiled about the catheter adjacent to the fluid connector or coiled within the fluid connector and in contact with fluid flowing through the fluid coupler, such as saline or blood. In some embodiments, the sensor can also include one or more additional working electrodes (e.g., for measuring baseline, for measuring a second analyte, or for measuring a substantially non-analyte related signal, and the like, such as described in more detail in U.S. Patent Publication No. US-2005-0143635-A1 and U.S. Patent Publication No. US-2007-0027385-A1. In some embodiments one or more counter electrodes can be provided on a surface of the catheter or within or on the fluid connector.

**[0328]** In some of the preferred embodiments, the catheter is designed to indwell within a host's blood flow (e.g., a peripheral vein or artery) and remain in the blood flow for a period of time (e.g., the catheter is not immediately removed). In some embodiments, the indwelling catheter can be inserted into the blood flow for example, for a few minutes or more, or from about 1 to 24 hours, or from about 1 to 10 days, or even longer. For example, the catheter can indwell in the host's blood stream during an entire perioperative period (e.g., from host admittance, through an operation, and to release from the hospital).

**[0329]** In some embodiments, the catheter is configured as an intravenous catheter (e.g., configured to be inserted into a vein). The catheter can be inserted into any commonly used vein, such as in a peripheral vein (e.g., one of the metacarpal veins of the arm); in some embodiments (e.g., such as described with reference to **Figs. 1A** to **1E**) the analyte sensor inserted into a catheter. In alternative embodiments, the sensor is integrally formed on a catheter such as described in more detail with reference to **Figs. 2A** to **2F,** for example. Other veins, such as leg or foot veins, hand veins, or even scalp or umbilical veins, can also be used.

**[0330]** In addition to sensing analyte levels *via* a sensor system as described herein, the intravenous catheter can be used for delivery of fluids and/or drugs to the host's circulatory system. The catheter can be configured to be coupled to other medical devices or functions, for example, saline, blood products, total parenteral feeding or medications can be given to the host *via* the indwelling intravenous catheter. In some embodiments, the catheter can be operatively connected to a pump, such as an infusion pump, to facilitate flow of the fluids into the host and a desired rate. For example, an infusion pump can pump saline into the host at a rate of 1cc per minute, or at higher or lower rates. The rate of infusion can be changed (increased or decreased). For example, an infusion can be temporarily stopped, to permit injection of pain medication into the IV system, followed by increasing the infusion rate (e.g., for 5 minutes) to rapidly deliver the pain medication to the host's circulatory system.

**[0331]** In some embodiments, the catheter is configured as an arterial catheter (e.g., configured to be inserted into an arterial line or as part of an arterial line). Typically, an arterial catheter is inserted in the wrist (radial artery), armpit (axillary artery), groin (femoral artery), or foot (pedal artery). Generally, arterial catheters provide access to the host's blood stream (arterial side) for removal of blood samples and/or application of test devices, such as but not limited to a pressure transducer (for measuring blood pressure automatically), however, arterial catheters can also be used for

delivery of fluids or medications. In one embodiment, a catheter is inserted into an arterial line and the sensor inserted into the catheter (e.g., functionally coupled) as described elsewhere herein. Saline filled non-compressible tubing is then coupled to the sensor, followed by a pressure transducer. An automatic flushing system (e.g., saline) is coupled to the tubing as well as a pressure bag to provide the necessary pressure. Electronics are generally operatively coupled to the pressure transducer for calculating and displaying a variety of parameters including blood pressure. Other medical devices can also be connected to the arterial catheter, to measure various blood components, such as but not limited to $O_2$, $CO_2$, $PCO_2$, $PO_2$, potassium, sodium, pH, lactate, urea, bilirubin, creatinine, hematocrit, various minerals, various metabolites, and the like.

**[0332]** In another embodiment, a blood pressure measurement system is inserted into the host and can be used as is known in the art. The analyte sensor (e.g., glucose sensor), such as the embodiment shown in **Figs. 1A-1E,** is inserted into the pre-inserted (e.g., already in-dwelling) catheter using the following general methodology. First, the pressure transducer is temporarily disabled by disconnecting from the pre-inserted catheter. A cap (optionally) covers the protective slotted sheath and can be removed so as to enable the sensor to be grasped at the fluid coupler. The sheath, which is generally more rigid than the sensor but less flexible than a needle, is then threaded through the pre-inserted catheter so as to extend beyond the catheter into the blood stream (e.g., by about 0.001 inches to about 1 inches). The sheath is then removed by sliding the sensor through a small outlet hole and/or slot in the sheath. Thus, the sensor remains within the pre-inserted catheter and the fluid coupler, which supports the distal portion of the sensor, is coupled to the catheter itself. Saline filled non-compressible tubing is then coupled to the second side (e.g., back end) of the fluid coupler. The sensor electronics (whether adjacent to the fluid coupler or otherwise wired to the fluid coupler) are then operatively connected (e.g., wired or wirelessly) to the sensor to initiate sensor function.

**[0333]** In some embodiments, a portion of the sensor system (e.g., sensor, catheter, or other component) can be configured to allow removal of blood samples from the host's blood stream (e.g., artery or vein). Sample removal can be done using any systems and methods known in the art, for example, as is practiced for removing a blood sample from an arterial catheter (e.g., and arterial line). In one such exemplary embodiment, any tubing or equipment coupled to the second side of the fluid coupler is disconnected. A syringe is then be coupled to the second side and blood removed *via* the catheter by pulling back on the syringe plunger. In a further embodiment, saline can be flushed through the fluid coupler and catheter. In another embodiment, the fluid coupler can be configured with a side valve, to allow coupling of a syringe, for removal of blood samples or delivery of fluids, such as medications, without disconnecting attached tubing of equipment, and the like. In still another embodiment, a valve or diaphragm, for access to the system by a syringe, can be coupled into the tubing at a short distance from the fluid coupler. In yet another embodiment, the sensor is integrally formed on the arterial catheter, such as the embodiment shown in **Figs. 2A-2B,** and tubing can be disconnected from the connector, a syringe operably associated with the connector, and blood removed with the syringe. After blood collection, the syringe is removed and the tubing reconnected to the connector.

**[0334]** In still another embodiment, the analyte sensor can be functionally coupled to an extracorporeal blood flow device. A variety of devices exist for testing various blood properties/analytes at the bedside, such as but not limited to the blood gas and chemistry devices manufactured by Via Medical, Austin, Texas, USA. These devices generally withdraw a blood sample from the host, test the blood sample, and then return it to the host. Such a device can be connected in series to the arterial catheter, with the sensor in-between, and using systems and methods known in the art. In one embodiment, a sensor, such as the embodiment shown in **Figs. 1A-1E,** is functionally connected to an in-dwelling arterial catheter, as described herein, and the extracorporeal blood flow device is connected to the second side of the fluid coupler. In an alternative embodiment, the sensor is integrally formed on the arterial catheter, such as the embodiment shown in **Figs. 2A-2F,** and the extracorporeal blood flow device is functionally connected to the connector **218.** Other devices, such as but not limited to dialysis machines, heart-lung bypass machines or blood collection bags, or other vascular access devices, can be functionally coupled to the analyte sensor.

**[0335]** The analyte sensor system of the preferred embodiments can be designed with a variety of alternative configurations. In some embodiments, the sensor is connected to a fluid connection device. The fluid connection device in these embodiments can be any standard fluid connection device known in the art, such as a fluid coupler, or a fluid coupler custom manufactured to preferred specifications. On its first side, the fluid coupler is configured to couple to an existing catheter or cannula (as described with reference to **Figs. 1A-1E**). The catheter (or cannula) is typically inserted into a vascular access device and/or into a hospital host during a hospital stay. For example, the catheter can be inserted into an arterial line (e.g., for removing blood samples or for measuring blood pressure using a pressure transducer) or a venous line (e.g., for intravenous delivery of drugs and other fluids). In general practice, the catheter is inserted into the host's blood vessel, for example, and maintained there for a period of time during the host's hospital stay, such as part of the stay or during the entire stay (e.g., perioperatively). In one alternative embodiment, another vascular access device (e.g., other than a catheter) can be used to receive the sensor. In yet another alternative embodiment, the sensor system of the preferred embodiments can be inserted into a vascular access device (e.g., rather than the vascular system directly). Some examples of vascular access devices include but are not limited to, catheters, shunts, automated blood withdrawal devices and the like.

**[0336]** In some embodiments, such as the embodiment illustrated in **Figs. 1A** to **1E,** the system **10** is configured such that the sensor is inserted into a vascular access device, such as but not limited to a catheter **12** (e.g., a catheter that has been inserted into the host's blood stream prior to sensor insertion). In general, catheters are small, flexible tubes (e.g., soft catheter) but they can also be larger, rigid tubes. Catheters are inserted into a host's body cavity, vessel, or duct to provide access for fluid removal or insertion, or for access to medical equipment. Catheters can also be inserted into extracorporeal devices, such as but not limed to an arterio-venous shunt for the transfer of blood from an artery to a vein. Some catheters are used to direct access to the circulatory system (e.g., venous or arterial catheters, Swan Gantz catheters) to allow removal of blood samples, the infusion of fluids (e.g., saline, medications, blood or total parenteral feeding) or access by medical devices (e.g., stents, extracorporeal blood chemistry analysis devices, invasive blood pressure monitors, etc.).

**[0337]** Preferably, the sensor is designed to include a protective cap, as illustrated in **Figs. 1A-1E.** Namely, **Fig. 1A** and **1B** illustrates the catheter (the catheter cap having been removed prior to insertion), well known to those skilled in the art, which can be inserted into the host's blood vessel using standard methods. The sensor **14** is configured for measurement of an analyte (e.g., glucose) in the host's body, and is in fluid connection within the catheter lumen, which is in fluid connection with the fluid coupler **20** of the sensor. The first side **20a** of the fluid coupler **20** of the sensor is designed to couple to the catheter, e.g., by screwing or snapping thereon, and can also couple (on its second side **20b**) with other medical devices. One advantage of the fluid coupler is that it provides for a small amount of bleed back, to prevent air bubbles in the host's blood stream.

**[0338]** The exemplary sensor system **10** of **Fig. 1A** and **1B** further includes a slotted protective sheath **26** that supports and protects the sensor during sensor insertion, for example, the sheath increases the sensor visibility (e.g., the sensor is so thin that it can be difficult for some people to see without the protective sheath) and provides for ease of sliding the sensor into the catheter. The slotted protective sheath is configured to fit within the fluid coupler and houses the sensor during insertion of the sensor into the catheter (e.g., an indwelling catheter within the host's blood flow). Preferably, the protective sheath is substantially more rigid than the sensor and at the same time substantially more flexible that a standard syringe needle, however other designs are possible. To facilitate removal of the protective sheath, a slot **30** is provided with an optional outlet hole **30a,** which is described in more detail with reference to **Fig. 1C**, and a hub **28.** By grasping and pulling the hub, the user (e.g., health care professional) can withdraw the protective sheath after coupling the fluid coupler to the catheter. Prior to insertion of the sensor, a cap is provided, to cover the protective sheath, for example, to keep the sheath and sensor sterile, and to prevent damage to the components during shipping and/or handling.

**[0339]** In general, the sensor system is configured with a potentiostat and/or sensor electronics that are operatively coupled to the sensor. In some embodiments, a portion of the sensor electronics, such as the potentiostat, can be disposed directly on the fluid coupler. However, some or all of the sensor electronics (including the potentiostat) can be disposed remotely from the fluid coupler (e.g., on the bedside or on a stand) and can be functionally coupled (e.g., wired or wireless), as is generally known to those skilled in the art.

**[0340]** **Figs. 1C1** and **1C2** are cross-sectional views (not to scale) of the fluid coupler, including a protective sheath **26,** a sensor **14,** and a cap **32** (cap to be removed prior to insertion) in one embodiment. The protective sheath **26** extends through the fluid coupler and houses the sensor, for sensor insertion into a catheter. The protective sheath includes an optional outlet hole **30a,** through which the sensor extends and a slot **30** along a length of the protective sheath that communicates with the outlet hole and enables the protective sheath to be removed after the sensor has been inserted into the host's body. The protective sheath includes a hub **28** for ease of handling.

**[0341]** In some embodiments, the glucose sensor is utilized in combination with another medical device (e.g., a medical device or access port that is already coupled to, applied to, or connected to the host) in a hospital or similar clinical setting. For example, a catheter can be inserted into the host's vein or artery, wherein the catheter can is connected to additional medical equipment. In an alternative example, the catheter is placed in the host to provide quick access to the host's circulatory system (in the event of a need arising) and is simply capped. In another example, a dialysis machine can be connected to the host's circulatory system. In another example, a central line can be connected to the host, for insertion of medical equipment at the heart (e.g., the medical equipment reaches the heart through the vascular system, from a peripheral location such as a leg or arm pit).

**[0342]** In practice of coupling to a catheter, before insertion of the sensor, the access port is opened. In one exemplary embodiment of a pre-inserted catheter that is capped, the cap is removed and the sensor inserted into the catheter. The back end of the sensor system can be capped or attached to additional medical equipment (e.g., saline drip, blood pressure transducer, dialysis machine, blood chemistry analysis device, etc.). In another exemplary embodiment, medical equipment (e.g., saline drip, blood pressure transducer, dialysis machine, blood chemistry analysis device, etc.) is already connected to the catheter. The medical equipment is disconnected from the catheter, the sensor inserted into (and coupled to) the catheter and then the medical equipment reconnected (e.g., coupled to the back end of the sensor system).

**[0343]** In some embodiments, the sensor is inserted directly into the host's circulatory system without a catheter or other medical device. In one such exemplary embodiment, the sheath covering the sensor is relatively rigid and supports

the sensor during insertion. After the sensor has been inserted into the host's vein or artery, the supportive sheath is removed, leaving the exposed sensor in the host's vein or artery. In an alternative example, the sensor is inserted into a vascular access device (e.g., with or without a catheter) and the sheath removed, to leave the sensor in the host's vein or artery (e.g., through the vascular access device).

**[0344]** In various embodiments, in practice, prior to insertion, the cap **32** over the protective sheath is removed as the health care professional holds the glucose sensor by the fluid coupler **20.** The protective sheath **26,** which is generally more rigid than the sensor but more flexible than a needle, is then threaded through the catheter so as to extend beyond the catheter into the blood flow (e.g., by about 0.010 inches to about 1 inches). The protective sheath is then removed by sliding the sensor through the (optional) outlet hole **30a** and slotted portion **30** of the sheath (e.g., by withdrawing the protective sheath by pulling the hub **28**). Thus the sensor remains within the catheter; and the fluid coupler **20,** which holds the sensor **14,** is coupled to the catheter itself (*via* its connector **18**). Other medical devices can be coupled to the second side of the fluid coupler as desired. The sensor electronics (e.g., adjacent to the fluid coupler or otherwise coupled to the fluid coupler) are then operatively connected (e.g., wired or wirelessly) to the sensor for proper sensor function as is known in the art.

**[0345]** In another embodiment, the catheter **12** includes a plurality of perforations (e.g., holes) that allow the host's fluid (e.g., blood) to flow through the lumen **12a** of the catheter. The fluid flowing through the catheter can make contact with a sensor **14** inserted therein. In a further embodiment, the sensor does not protrude out of the catheter's tip **12b** and the host's blood flowing through the perforated catheter's lumen contacts the sensor's electroactive surfaces.

**[0346]** In still another embodiment, the catheter **12** includes at least a first lumen and a second lumen. The sensor **14** is configured for insertion into the catheter's first lumen. The second lumen can be used for infusions into the host's circulatory system or sample removal without disturbing the sensor within the first lumen.

**[0347]** **Figs. 2A-2F** are schematic views of a sensor integrally formed (integrally incorporated) onto a surface of a catheter, in some exemplary embodiments. In some embodiments, the sensor can be integrally formed on an exterior surface **232** of the catheter. In other embodiments, the sensor can be integrally formed on an interior surface of the catheter (e.g., on a lumenal surface). In still other embodiments, the sensor can be integrally formed on the sensor's tip (e.g., as indicated by **214a**). In yet other embodiments, the sensor can be integrally incorporated with the catheter, for example by bonding a sensor of the type described in **Figs. 3A** to **3C** into an inner or outer surface of the catheter.

**[0348]** In some embodiments, one or more of the electrodes is deposited on the *in vivo* portion of the catheter **212,** such as *via* screen-printing and/or electrospinning. In some embodiments, at least one of the electrodes **240,** such as but not limited to a counter and/or a reference electrode is deposited within the *ex vivo* portion of the catheter (e.g., within the connector/hub). In one embodiment, two working electrodes **240** are disposed on the exterior surface **232** of the catheter's *in vivo* portion. The first working electrode is configured to generate a signal associated with the analyte and with non-analyte-related species that have an oxidation/reduction potential that overlaps with that of the analyte. The second working electrode is configured to generate a signal associated with non-analyte-related species that have an oxidation/reduction potential that overlaps with that of the analyte. As described elsewhere herein, the signals of the first and second working electrodes can be processed to provide a substantially analyte-only signal. Continuous analyte sensors comprising two working electrodes are described in greater detail elsewhere herein, in U.S. Patent Publication Nos. US-2007-0027385-A1, US-2007-0213611-A1, US-2007-0027284-A1, US-2007-0032717-A1, and US-2007-0093704, and U.S. Patent Application No. 11/865,572 filed on October 1, 2007 and entitled "DUAL-ELECTRODE SYSTEM FOR A CONTINUOUS ANALYTE SENSOR,"

**[0349]** In some alternative embodiments, one or more analyte sensors are disposed (e.g., deposited, formed) on the exterior surface of the *in vivo* portion of the catheter. Each sensor can include one, two or more working electrodes. The electrodes can be configured as described elsewhere therein. In some embodiments, the catheter **12** is configured with two or more analyte sensors, wherein each of the sensor is configured to detect a different analyte and/or a property of the sample, as described elsewhere herein. For example, in some embodiments, the sensors are configured to detect at least two analytes such as but not limited to glucose, $O_2$, $CO_2$, potassium, sodium, $H^+$, $OH^-$, lactate, urea, bilirubin, creatinine, various minerals, various metabolites, and the like. In some embodiments, at least one of the sensors is configured to detect a property of the host's blood, such as but not limited to pH, oxygen tension, $PCO_2$, $PO_2$, temperature, hematocrit, and the like. In some circumstances, one or more of the plurality of analyte sensor can be configured as a back-up or redundant sensor to a first sensor, such as to confirm the correct functioning of the first sensor. For example, two glucose sensor could be disposed within the connector, such that the second glucose sensor provides a confirmation of the first glucose sensor's measurements.

**[0350]** Generally, the sensor system is provided with a cap that covers the catheter and *in vivo* portion of the integral sensor. A needle or trochar that runs the length of the catheter supports the device during insertion into the host's blood stream. Prior to use, medical caregiver holds the device by the fluid connector **218** and removes the cap to expose the *in vivo* portion of the device (e.g., the catheter). The caregiver inserts the *in vivo* portion of the device into one of the host's veins or arteries (depending upon whether the catheter is an intravenous catheter or an arterial catheter). After insertion, the needle is withdrawn from the device. The device is then capped or connected to other medical equipment

(e.g., saline bag, pressure transducer, blood collection bag, total parenteral feeding, dialysis equipment, automated blood chemistry equipment, etc.). In some alternative embodiments, the sensor-integrated catheter can be in communication (e.g., fluid communication) with the host's vascular system through a vascular access device.

**[0351]** In some embodiments, a glucose sensor system includes a sensing mechanism substantially similar to that described in U.S. Patent Publication No. US-2006-0020187-A1; for example, with platinum working electrode and silver reference electrode coiled there around. Alternatively, the reference electrode can be located remote from the working electrode so as not to be inserted into the host, and can be located, for example, within the fluid coupler, thereby allowing a smaller footprint in the portion of the sensor adapted for insertion into the body (e.g., blood stream); for example, without a coiled or otherwise configured reference electrode proximal to the working electrode. Although a platinum working electrode is discussed, a variety of known working electrode materials can be utilized (e.g., Platinum-Iridium or Iridium). When located remotely, the reference electrode can be located away from the working electrode (e.g., the electroactive portion) at any location and with any configuration so as to maintain bodily and/or in fluid communication therewith as is appreciated by one skilled in the art.

**[0352]** In an alternative embodiment, the sensor tip **14a** includes an enlarged, atraumatic area, for example a dull or bulbous portion about two times the diameter of the sensor or larger. In one exemplary embodiment, the enlarged portion is created by heating, welding, crushing or bonding a substantially rounded structure onto the tip of the sensor (e.g., polymer or metal). In another exemplary embodiment, the tip of the sensor is heated (e.g., arc welded or flash-butt resistance welded) to cause the tip to enlarge (e.g., by melting). The enlarged portion can be of any atraumatic shape, such as but not limited to oval, round, cone-shaped, cylindrical, teardrop, etc. While not wishing to be bound by theory, it is believed that an atraumatic or enlarged area enables enhanced stability of a small diameter sensor in the blood flow and ensures that the sensor remains within the blood flow (e.g., to avoid piercing a vessel wall and/or becoming inserted subluminally.)

**[0353]** In some embodiments, a second working electrode can be provided on the sensor for measuring baseline, and thereby subtracting the baseline from the first working electrode to obtain a glucose-only signal, as disclosed in copending U.S. Patent Publication No. US-2005-0143635-A1 and U.S. Patent Publication No. US-2007-0027385-A1.

**[0354]** Referring now to **Figs. 2A-2E** in more detail, some embodiments of the analyte sensor system include a catheter **212** adapted for inserting into a host in a hospital or clinical setting, wherein the analyte sensor **214** is built integrally with the catheter **212.** For example, a glucose sensor can be integrally formed on the catheter itself. **Figs 2A-2B** illustrate one embodiment, wherein the catheter **212** is configured both for insertion into a host, and can be configured to couple to other medical devices on its *ex vivo* end. However, coupling to other medical devices is not necessary. In some embodiments, the catheter includes a connector **218** configured for connection to tubing or other medical devices, as described herein. The embodiment shown in **Figs. 2A-2B** includes two or three electrodes **240** on the outer surface of the *in vivo* portion of the catheter **212.** In some embodiments, the catheter is perforated (as described elsewhere herein) and at least one electrode is disposed within the lumen (not shown) of the perforated catheter. In some embodiments, the catheter includes a single lumen. In other embodiment, the catheter includes two or more lumens.

**[0355]** With reference to **Figs. 2C-2E,** in some embodiments, at least one working electrode **240** is disposed on the exterior surface of the *in vivo* portion of the catheter. Alternatively, the at least one working electrode can be disposed on an interior surface of the catheter, the tip of the catheter, extend from the catheter, and the like. In general, the preferred embodiments can be designed with any number of electrodes, including one or more counter electrodes, one or more reference electrodes, and/or one or more auxiliary working electrodes. In further embodiments, the electrodes can be of relatively larger or smaller surface area, depending upon their uses. In one example, a sensor includes a working electrode and a reference electrode that has a larger surface area (relative to the surface area of the working electrode) on the surface of the catheter. In another example, a sensor includes a working electrode, a counter electrode, and a reference electrode sized to have an increased surface area as compared to the working and/or counter electrode. In some embodiments, the reference electrode is disposed at a location remote from the working electrode, such as within the connector (e.g., coiled within the connector). In some embodiments, the reference electrode is located on the host's body (e.g., in body contact).

**[0356]** The electrodes **240** can be deposited on the catheter using any suitable techniques known in the art, for example, thick or thin film deposition techniques. The electrodes can be formed of any advantageous electrode materials known in the art (e.g., platinum, platinum-iridium, palladium, graphite, gold, carbon, silver, silver-silver chloride, conductive polymer, alloys, combinations thereof, and the like). In other embodiments, one or more of the electrodes is formed from an electrically conductive material (e.g., wire or foil comprising platinum, platinum-iridium, palladium, graphite, gold, carbon, silver, silver-silver chloride, conductive polymer, alloys, combinations thereof, and the like) applied to the exterior surface of the catheter, such as but not limited twisting, coiling, rolling or adhering.

**[0357]** In some embodiments, the catheter is (wired or wirelessly) connected to sensor electronics (not shown, disposed on the catheter's connector and/or remote from the catheter) so as to electrically connect the electrodes on the catheter with the sensor electronics. The inserted catheter (including the sensor integrally formed thereon) can be utilized by other medical devices for a variety of functions (e.g., blood pressure monitor, drug delivery, etc).

**[0358]** Referring now to **Figs. 2G** through **2L,** in some preferred embodiments a plurality of analyte sensors **240** are disposed within a widened portion of the catheter, such as but not limited to a flared portion and/or a connector portion **212,** or within the interior of a connector **250,** such as but not limited to a Leur lock, a Y-connector, a T-connector, an X-connector, and a valve, wherein a first end of the connector is configured to be coupled/connected to another vascular access device, such as a catheter or cannula, and a second end of the connector is configured to be coupled/connected to other IV equipment, such as another connector, a valve, IV tubing, and the like.

**[0359]** **Fig. 2G** is a cross section of a vascular access device comprising a plurality of analyte sensors **240** in one embodiment. At the proximal end, also referred to herein as the *in vivo* portion, the vascular access device includes a catheter **212** having a lumen **212a** and a small orifice **212b.** At the distal end, also referred to here as the *ex vivo* portion, the vascular access device includes a connector **218,** also referred to herein as the "hub." The hub includes an orifice **218c,** which is configured for connection (e.g., fluid communication) with other IV equipment, such as *via* one or more flanges **218a.** The connector **218** also includes a duct **218b,** also referred to a widened portion (as compared to lumen **212a,** which may be referred to as the connector's lumen. A plurality of analyte sensors **240** is disposed within the duct **218b.** In some embodiments, one or more of the analyte sensors **240** is deposited/formed on a surface of the duct **218b,** such as by silk screening or other useful deposition techniques. In some embodiments, one or more of the analyte sensors **240** is applied to the duct's surface, such as by adhering or micro welding a previously formed sensor to the duct's surface. In some embodiments, at least a portion of one or more of the analyte sensors **240** is unattached to the duct's surface. The analyte sensors can be configured to detect one or more analytes using any means known in the art, such as but not limited to electrochemical detection, enzymatic detection, chemical detection, physical detection, immunochemical detection, optical detection, radiometric detection, and combinations thereof. For example, in one embodiment of a device including three sensors **240** within the hub **218,** a first sensor **240** can be configured to detect glucose electrochemically, a second sensor **240** can be configured to detect oxygen optically, and the third sensor **240** can be configured to detect bilirubin immunochemically.

**[0360]** **Fig. 2H** is a cross section of a vascular access device comprising a plurality of analyte sensor **240** in another embodiment. In this embodiment, the vascular access device is a connector **250** and/or valve, such as but not limited to a Leur lock, a Leur lock, a Y-connector, a T-connector, and an X-connector. In general, the connector **250** is configured to be coupled/connected to vascular access devices, such that a fluid can pass between two vascular access devices coupled to the connector's two ends. For example, a first end of the connector can be coupled to a catheter or cannula implanted (e.g., pre-implanted) in a host's vein or artery, and a second end of the connector can be coupled to another connector, a valve, IV tubing, and IV bag, a test device, etc. The connector includes a lumen/duct **254** and a proximal orifice **258.** A plurality of analyte sensors **240** are disposed within the duct **254.** As described with reference to the device shown in **Fig. 2G,** the plurality of analyte sensors can be disposed within the duct **254** using any means known in the art. In some embodiments, one or more of the analyte sensors are deposited (e.g., formed on) the surface of the duct **254.** In some embodiments, one or more of the analyte sensors are applied to the surface of the duct **254.** In some embodiments, one or more of the analyte sensors is configured to pass through the wall **252** of the connector such that a first portion of the sensor **240** is disposed within the duct **256** and a second portion of the sensor **240** is disposed at the exterior of the connector **250** (described in more detail herein).

**[0361]** **Fig. 2I** is a cross-section of a vascular access device of either **Fig. 2G** or **Fig. 2H** taken along line **2I - 2I,** looking towards the proximal end of the vascular access device. The device includes a duct/lumen **212b/254** defined by a wall **260.** The *in vivo* orifice (also referred to as the proximal orifice with relation to the host) of the device is represented by circle **212b/258.** As shown in this embodiment, a plurality of sensors can be disposed within the duct, such as but not limited at the in the interior surface of the wall. In some embodiments, the device includes two analyte sensors. In some embodiments, the device includes 3, 4, 5, 6, 7 or more analyte sensors. In some embodiments, one or more of the analyte sensors are configured to be disposed entirely within the duct (e.g., to not protrude out of the duct). In some embodiments, one or more analyte sensors can be configured such that a portion thereof protrudes out the duct, such as but not limited to into the lumen of a catheter **212** or through the proximal orifice **212b/258** of the device. In some embodiments, a portion or one or more of the sensors can be configured to protrude through the *ex vivo* orifice (also referred to as the distal orifice with ration to the host) of the device. The analyte sensors 240 disposed within the device can be of any configuration and can use any detection method, including but not limited to electrochemical, enzymatic, optical, radiometric, chemical, physical, immunochemical and the like, including a combination thereof.

**[0362]** **Fig. 2J** is a cross-section of a vascular access device of either **Fig. 2G** or **Fig. 2H** taken along line **2I - 2I,** looking towards the proximal end of the vascular access device, prior to installation of any analyte sensors **240. Fig. 2K** depicts the **Fig. 2J** device after sensor installation. In this embodiment, a plurality of sensor sites **262** is located at the surface of the wall **260.** While **Figs. 2J** and **2K** depict the sensor sites **262** as being depressions in the wall **260,** the sensor sites **262** can be of any configuration, such as but not limited to a portion of the wall's inner surface that is flush with the remaining portion of the inner surface, a textured portion of the inner surface, a channel, a hole, and the like. In some embodiments, the sensor sites can have a plurality of configurations. For example, in a device including four sensor sited **262,** a first site can have a first configuration, the second and third sites a second configuration, and the

fourth site yet another configuration.

**[0363]** **Fig. 2L** is a cross-section of a vascular access device of either **Fig. 2G** or **Fig. 2H** taken along line **2I** - **2I**, looking towards the proximal end of the vascular access device, in an alternative embodiment. In this embodiment, the sensor sites **262** can be formed to include a plug **264** and/or a breakaway portion of the wall **260,** which can be removed to enable sensor installation. For example, a plug/breakaway portion can be pushed and/or punched out of the sensor site and then the sensor installed in the sensor site. In some embodiments, removal of a plug/breakaway portion creates a channel through the wall, such that a sensor (at least a portion thereof) can be inserted through the channel and into the duct **254.** In some embodiments, the portion of an installed sensor remaining on the external side of the wall is configured to functionally connect to sensor electronics, as is appreciated by one skilled in the art. While not wishing to be bound by theory, it is believed that this configuration enables increased accuracy and speed in device assembly because the sensors can be manufactured separately from the device and then installed into the device in a "plug-and-play" fashion.

**[0364]** In some embodiments, the device is formed by injection molding, using techniques known in the art. In one exemplary embodiment, the sensors are placed in a mold, which is configured to hold the sensors in such an orientation that after the injection molding procedure, the sensors will be in the correct location and/or orientation for correct function of the device. After the sensors are placed in the mold, the mold is closed and injected with a material (e.g., molten plastic). During the injection molding process, the wall **260** of the device is thus formed about a portion of each sensor 240, such that a sensing portion of each sensor will be disposed within the duct 212b/258 and another portion of each sensor (e.g., a portion configured for connection to sensor electronics) will be disposed at the exterior of the device. Similar manufacturing techniques are used for the manufacture of syringes and lancets, wherein the plastic portion of the device is formed about a portion of the needle.

**[0365]** While not wishing to be bound by theory, a number of the systems and methods disclosed in the preferred embodiments (e.g., an analyte sensor to be disposed in communication with the host's blood), can be employed in transcutaneous (e.g., transdermal) or wholly implantable analyte sensor devices. For example, the sensor could be integrally formed on the *in vivo* portion of a subcutaneous device or a wholly implantable device. As another example, an enlarged surface area (e.g., bulbous end) can useful in the design of a transcutaneous analyte sensor.

Exemplary Sensor Configurations

**[0366]** Referring to **Figs. 3A** to **3C,** in some embodiments, the sensor can be configured similarly to the continuous analyte sensors disclosed in co-pending U.S. Patent Publication No. US-2007-0197889-A1. The sensor includes a distal portion **342,** also referred to as the *in vivo* portion, adapted for insertion into the catheter as described above, and a proximal portion **340,** also referred to as an *ex vivo* portion, adapted to operably connect to the sensor electronics. Preferably, the sensor includes two or more electrodes: a working electrode **344** and at least one additional electrode, which can function as a counter electrode and/or reference electrode, hereinafter referred to as the reference electrode **346.** A membrane system is preferably deposited over the electrodes, such as described in more detail with reference to **Figs. 3A** to **3C,** below.

**[0367]** **Fig. 3B** is an expanded cutaway view of a distal portion of the sensor in one embodiment, showing working and reference electrodes. In preferred embodiments, the sensor is formed from a working electrode **344** (e.g., a wire) and a reference electrode **346** helically wound around the working electrode **344.** An insulator **345** is disposed between the working and reference electrodes to provide electrical insulation therebetween. Certain portions of the electrodes are exposed to enable electrochemical reaction thereon, for example, a window **343** can be formed in the insulator to expose a portion of the working electrode **344** for electrochemical reaction.

**[0368]** In preferred embodiments, each electrode is formed from a fine wire with a diameter of from about 0.001 inches or less to about 0.050 inches or more, for example, and is formed from, *e.g.*, a plated insulator, a plated wire, or bulk electrically conductive material. For example, in some embodiments, the wire used to form a working electrode is about 0.002, 0.003, 0.004, 0.005, 0.006, 0.007, 0.008, 0.009, 0.010, 0.015, 0.020, 0.025, 0.030, 0.035, 0.040 or 0.045 inches in diameter. Although the illustrated electrode configuration and associated text describe one preferred method for forming a sensor, a variety of known sensor configurations can be employed with the analyte sensor system of the preferred embodiments, such as U.S. Patent No. 5,711,861 to Ward et al., U.S. Patent No. 6,642,015 to Vachon et al., U.S. Patent No. 6,654,625 to Say et al., U.S. Patent 6,565,509 to Say et al., U.S. Patent No. 6,514,718 to Heller, U.S. Patent No. 6,465,066 to Essenpreis et al., U.S. Patent No. 6,214,185 to Offenbacher et al., U.S. Patent No. 5,310,469 to Cunningham et al., and U.S. Patent No. 5,683,562 to Shaffer et al., U.S. Patent 6,579,690 to Bonnecaze et al., U.S. Patent 6,484,046 to Say et al., U.S. Patent 6,512,939 to Colvin et al., U.S. Patent 6,424,847 to Mastrototaro et al., U.S. Patent 6,424,847 to Mastrototaro et al, for example.

**[0369]** The above patents are not inclusive of all applicable analyte sensors; in general, it should be understood that the disclosed embodiments are applicable to a variety of analyte sensor configurations. It is noted that much of the description of the preferred embodiments, for example the membrane system described below, can be implemented

not only with *in vivo* sensors, but also with *in vitro* sensors, such as blood glucose meters (SMBG).

**[0370]** In some embodiments, the working electrode comprises a wire formed from a conductive material, such as platinum, platinum-iridium, palladium, graphite, gold, carbon, conductive polymer, alloys, and the like. Although the electrodes can by formed by a variety of manufacturing techniques (bulk metal processing, deposition of metal onto a substrate, and the like), it can be advantageous to form the electrodes from plated wire (*e.g.*, platinum on steel wire) or bulk metal (*e.g.*, platinum wire). It is believed that electrodes formed from bulk metal wire provide superior performance (*e.g.*, in contrast to deposited electrodes), including increased stability of assay, simplified manufacturability, resistance to contamination (*e.g.*, which can be introduced in deposition processes), and improved surface reaction (*e.g.*, due to purity of material) without peeling or delamination.

**[0371]** In some embodiments, the working electrode is formed of platinum-iridium or iridium wire. In general, platinum-iridium and iridium materials are generally stronger (e.g., more resilient and less likely to fail due to stress or strain fracture or fatigue). It is believed that platinum-iridium and/or iridium materials can facilitate a wire with a smaller diameter to further decrease the maximum diameter (size) of the sensor (e.g., *in vivo* portion). Advantageously, a smaller sensor diameter both reduces the risk of clot or thrombus formation (or other foreign body response) and allows the use of smaller catheters.

**[0372]** The electroactive window **343** of the working electrode **344** is configured to measure the concentration of an analyte. In an enzymatic electrochemical sensor for detecting glucose, for example, the working electrode measures the hydrogen peroxide produced by an enzyme catalyzed reaction of the analyte being detected and creates a measurable electronic current For example, in the detection of glucose wherein glucose oxidase produces hydrogen peroxide as a byproduct, hydrogen peroxide reacts with the surface of the working electrode producing two protons ($2H^+$), two electrons ($2e^-$) and one molecule of oxygen ($O_2$), which produces the electronic current being detected.

**[0373]** In preferred embodiments, the working electrode **344** is covered with an insulating material **345,** for example, a non-conductive polymer. Dip-coating, spray-coating, vapor-deposition, or other coating or deposition techniques can be used to deposit the insulating material on the working electrode. In one embodiment, the insulating material comprises parylene, which can be an advantageous polymer coating for its strength, lubricity, and electrical insulation properties. Generally, parylene is produced by vapor deposition and polymerization of para-xylylene (or its substituted derivatives). While not wishing to be bound by theory, it is believed that the lubricious (e.g., smooth) coating (e.g., parylene) on the sensors of some embodiments contributes to minimal trauma and extended sensor life. While parylene coatings are generally preferred in some embodiments, any suitable insulating material can be used, for example, fluorinated polymers, polyethyleneterephthalate, polyurethane, polyimide, other nonconducting polymers, and the like. Glass or ceramic materials can also be employed. Other materials suitable for use include surface energy modified coating systems such as are marketed under the trade names AMC 18, AMC148, AMC141, and AMC321 by Advanced Materials Components Express of Bellafonte, PA. In some alternative embodiments, however, the working electrode may not require a coating of insulator.

**[0374]** The reference electrode **346,** which can function as a reference electrode alone, or as a dual reference and counter electrode, is formed from silver, silver/silver chloride, and the like. In some embodiments, the reference electrode **346** is juxtapositioned and/or twisted with or around the working electrode **344;** however other configurations are also possible (*e.g.*, coiled within the fluid connector, or an intradermal or on-skin reference electrode). In the illustrated embodiments, the reference electrode **346** is helically wound around the working electrode **344.** The assembly of wires is then optionally coated or adhered together with an insulating material, similar to that described above, so as to provide an insulating attachment.

**[0375]** In some embodiments, a silver wire is formed onto the sensor as described above, and subsequently chloridized to form silver/silver chloride reference electrode. Advantageously, chloridizing the silver wire as described herein enables the manufacture of a reference electrode with optimal *in vivo* performance. Namely, by controlling the quantity and amount of chloridization of the silver to form silver/silver chloride, improved break-in time, stability of the reference electrode, and extended life has been shown with some embodiments. Additionally, use of silver chloride as described above allows for relatively inexpensive and simple manufacture of the reference electrode.

**[0376]** In embodiments wherein an outer insulator is disposed, a portion of the coated assembly structure can be stripped or otherwise removed, for example, by hand, excimer lasing, chemical etching, laser ablation, grit-blasting (*e.g.*, with sodium bicarbonate or other suitable grit), and the like, to expose the electroactive surfaces. Alternatively, a portion of the electrode can be masked prior to depositing the insulator in order to maintain an exposed electroactive surface area. In one exemplary embodiment, grit blasting is implemented to expose the electroactive surfaces, preferably utilizing a grit material that is sufficiently hard to ablate the polymer material, while being sufficiently soft so as to minimize or avoid damage to the underlying metal electrode (*e.g.*, a platinum electrode). Although a variety of "grit" materials can be used (*e.g.*, sand, talc, walnut shell, ground plastic, sea salt, and the like), in some preferred embodiments, sodium bicarbonate is an advantageous grit-material because it is sufficiently hard to ablate, *e.g.*, a parylene coating, without damaging, *e.g.*, an underlying platinum conductor. One additional advantage of sodium bicarbonate blasting includes its polishing action on the metal as it strips the polymer layer, thereby eliminating a cleaning step that might otherwise

be necessary.

**[0377]** In the embodiment illustrated in **Fig. 3B,** a radial window **343** is formed through the insulating material **345** to expose a circumferential electroactive surface of the working electrode. Additionally, sections of electroactive surface of the reference electrode are exposed. For example, the sections of electroactive surface can be masked during deposition of an outer insulating layer or etched after deposition of an outer insulating layer.

**[0378]** In some applications, cellular attack or migration of cells to the sensor can cause reduced sensitivity and/or function of the device, particularly after the first day of implantation. However, when the exposed electroactive surface is distributed circumferentially about the sensor (*e.g.*, as in a radial window), the available surface area for reaction can be sufficiently distributed so as to minimize the effect of local cellular invasion of the sensor on the sensor signal. Alternatively, a tangential exposed electroactive window can be formed, for example, by stripping only one side of the coated assembly structure. In other alternative embodiments, the window can be provided at the tip of the coated assembly structure such that the electroactive surfaces are exposed at the tip of the sensor. Other methods and configurations for exposing electroactive surfaces can also be employed.

**[0379]** In some embodiments, the working electrode has a diameter of from about 0.001 inches or less to about 0.010 inches or more, preferably from about 0.002 inches to about 0.008 inches, and more preferably from about 0.004 inches to about 0.005 inches. The length of the window can be from about 0.1 mm (about 0.004 inches) or less to about 2 mm (about 0.078 inches) or more, and preferably from about 0.25 mm (about 0.01 inches) to about 0.375 mm (about 0.015 inches). In such embodiments, the exposed surface area of the working electrode is preferably from about 0.000013 $in^2$ (0.0000839$cm^2$) or less to about 0.0025 $in^2$ (0.016129 $cm^2$) or more (assuming a diameter of from about 0.001 inches to about 0.010 inches and a length of from about 0.004 inches to about 0.078 inches). The preferred exposed surface area of the working electrode is selected to produce an analyte signal with a current in the picoAmp range, such as is described in more detail elsewhere herein. However, a current in the picoAmp range can be dependent upon a variety of factors, for example the electronic circuitry design (*e.g.*, sample rate, current draw, A/D converter bit resolution, etc.), the membrane system (*e.g.*, permeability of the analyte through the membrane system), and the exposed surface area of the working electrode. Accordingly, the exposed electroactive working electrode surface area can be selected to have a value greater than or less than the above-described ranges taking into consideration alterations in the membrane system and/or electronic circuitry. In preferred embodiments of a glucose sensor, it can be advantageous to minimize the surface area of the working electrode while maximizing the diffusivity of glucose in order to optimize the signal-to-noise ratio while maintaining sensor performance in both high and low glucose concentration ranges.

**[0380]** In some alternative embodiments, the exposed surface area of the working (and/or other) electrode can be increased by altering the cross-section of the electrode itself. For example, in some embodiments the cross-section of the working electrode can be defined by a cross, star, cloverleaf, ribbed, dimpled, ridged, irregular, or other non-circular configuration; thus, for any predetermined length of electrode, a specific increased surface area can be achieved (as compared to the area achieved by a circular cross-section). Increasing the surface area of the working electrode can be advantageous in providing an increased signal responsive to the analyte concentration, which in turn can be helpful in improving the signal-to-noise ratio, for example.

**[0381]** In some alternative embodiments, additional electrodes can be included within the assembly, for example, a three-electrode system (working, reference, and counter electrodes) and/or an additional working electrode (*e.g.*, an electrode which can be used to generate oxygen, which is configured as a baseline subtracting electrode, or which is configured for measuring additional analytes). U.S. Patent Publication No. US-2005-0161346-A1, U.S. Patent Publication No. US-2005-0143635-A1, and U.S. Patent Publication No. US-2007-0027385-A1 describe some systems and methods for implementing and using additional working, counter, and/or reference electrodes. In one implementation wherein the sensor comprises two working electrodes, the two working electrodes are juxtapositioned (*e.g.*, extend parallel to each other), around which the reference electrode is disposed (e.g., helically wound). In some embodiments wherein two or more working electrodes are provided, the working electrodes can be formed in a double-, triple-, quad-, etc. helix configuration along the length of the sensor (for example, surrounding a reference electrode, insulated rod, or other support structure). The resulting electrode system can be configured with an appropriate membrane system, wherein the first working electrode is configured to measure a first signal comprising glucose and baseline (e.g., background noise) and the additional working electrode is configured to measure a baseline signal consisting of baseline only (*e.g.*, configured to be substantially similar to the first working electrode without an enzyme disposed thereon). In this way, the baseline signal can be subtracted from the first signal to produce a glucose-only signal that is substantially not subject to fluctuations in the baseline and/or interfering species on the signal.

**[0382]** Although the embodiments of **Figs. 3A** to **3C** illustrate one electrode configuration including one bulk metal wire helically wound around another bulk metal wire, other electrode configurations are also contemplated. In an alternative embodiment, the working electrode comprises a tube with a reference electrode disposed or coiled inside, including an insulator therebetween. Alternatively, the reference electrode comprises a tube with a working electrode disposed or coiled inside, including an insulator therebetween. In another alternative embodiment, a polymer (*e.g.*, insulating) rod is provided, wherein the electrodes are deposited (*e.g.*, electro-plated) thereon. In yet another alternative embodiment,

a metallic (*e.g.*, steel) rod is provided, coated with an insulating material, onto which the working and reference electrodes are deposited. In yet another alternative embodiment, one or more working electrodes are helically wound around a reference electrode.

**[0383]** Preferably, the electrodes and membrane systems of the preferred embodiments are coaxially formed, namely, the electrodes and/or membrane system all share the same central axle. While not wishing to be bound by theory, it is believed that a coaxial design of the sensor enables a symmetrical design without a preferred bend radius. Namely, in contrast to prior art sensors comprising a substantially planar configuration that can suffer from regular bending about the plane of the sensor, the coaxial design of the preferred embodiments do not have a preferred bend radius and therefore are not subject to regular bending about a particular plane (which can cause fatigue failures and the like). However, non-coaxial sensors can be implemented with the sensor system of the preferred embodiments.

**[0384]** In addition to the above-described advantages, the coaxial sensor design of the preferred embodiments enables the diameter of the connecting end of the sensor (proximal portion) to be substantially the same as that of the sensing end (distal portion) such that the protective slotted sheath is able to insert the sensor into the catheter and subsequently slide back over the sensor and release the sensor from the protective slotted sheath, without complex multi-component designs.

**[0385]** In one such alternative embodiment, the two wires of the sensor are held apart and configured for insertion into the catheter in proximal but separate locations. The separation of the working and reference electrodes in such an embodiment can provide additional electrochemical stability with simplified manufacture and electrical connectivity. One skilled in the art will appreciate that a variety of electrode configurations can be implemented with the preferred embodiments.

**[0386]** In addition to the above-described configurations, the reference electrode can be separated from the working electrode, and coiled within a portion of the fluid connector, in some embodiments. In another embodiment, the reference electrode is coiled within the fluid connector and adjacent to its first side. In an alterative embodiment, the reference electrode is coiled within the fluid connector and adjacent to its second side. In such embodiments, the reference electrode is in contact with fluid, such as saline from a saline drip that is flowing into the host, or such as blood that is being withdrawn from the host. While not wishing to be bound by theory, this configuration is believed to be advantageous because the sensor is thinner, allowing the use of smaller catheters and/or a reduced likelihood to thrombus production.

**[0387]** In another embodiment, the reference electrode **346** can be disposed farther away from the electroactive portion of the working electrode **343** (e.g., closer to the fluid connector). In some embodiments, the reference electrode is located proximal to or within the fluid coupler, such as but not limited to, coiled about the catheter adjacent to the fluid coupler or coiled within the fluid coupler and in contact with fluid flowing through the fluid coupler, such as saline. These configurations can also minimize at least a portion of the sensor diameter and thereby allow the use of smaller catheters and reduce the risk of clots.

**[0388]** In addition to the embodiments described above, the sensor can be configured with additional working electrodes as described in U.S. Patent Publication No. US-2005-0143635-A1, U.S. Patent No. 7,081,195, and U.S. Patent Publication No. US-2007-0027385-A1. For example, in one embodiment have an auxiliary working electrode, wherein the auxiliary working electrode comprises a wire formed from a conductive material, such as described with reference to the glucose-measuring working electrode above. Preferably, the reference electrode, which can function as a reference electrode alone, or as a dual reference and counter electrode, is formed from silver, Silver/Silver chloride, and the like.

**[0389]** In some embodiments, the electrodes are juxtapositioned and/or twisted with or around each other; however other configurations are also possible. In one example, the auxiliary working electrode and reference electrode can be helically wound around the glucose-measuring working electrode. Alternatively, the auxiliary working electrode and reference electrode can be formed as a double helix around a length of the glucose-measuring working electrode. The assembly of wires can then be optionally coated together with an insulating material, similar to that described above, in order to provide an insulating attachment. Some portion of the coated assembly structure is then stripped, for example using an excimer laser, chemical etching, and the like, to expose the necessary electroactive surfaces. In some alternative embodiments, additional electrodes can be included within the assembly, for example, a three-electrode system (including separate reference and counter electrodes) as is appreciated by one skilled in the art.

**[0390]** In some alternative embodiments, the sensor is configured as a dual-electrode system. In one such dual-electrode system, a first electrode functions as a hydrogen peroxide sensor including a membrane system containing glucose-oxidase disposed thereon, which operates as described herein. A second electrode is a hydrogen peroxide sensor that is configured similar to the first electrode, but with a modified membrane system (without active enzyme, for example). This second electrode provides a signal composed mostly of the baseline signal, b.

**[0391]** In some dual-electrode systems, the baseline signal is (electronically or digitally) subtracted from the glucose signal to obtain a glucose signal substantially without baseline. Accordingly, calibration of the resultant difference signal can be performed by solving the equation $y=mx$ with a single paired measurement. Calibration of the inserted sensor in this alternative embodiment can be made less dependent on the values/range of the paired measurements, less sensitive to error in manual blood glucose measurements, and can facilitate the sensor's use as a primary source of glucose

information for the user. U.S. Patent Publication No. US-2005-0143635-A1 describes systems and methods for subtracting the baseline from a sensor signal.

**[0392]** In some alternative dual-electrode system embodiments, the analyte sensor is configured to transmit signals obtained from each electrode separately (e.g., without subtraction of the baseline signal). In this way, the receiver can process these signals to determine additional information about the sensor and/or analyte concentration. For example, by comparing the signals from the first and second electrodes, changes in baseline and/or sensitivity can be detected and/or measured and used to update calibration (e.g., without the use of a reference analyte value). In one such example, by monitoring the corresponding first and second signals over time, an amount of signal contributed by baseline can be measured. In another such example, by comparing fluctuations in the correlating signals over time, changes in sensitivity can be detected and/or measured.

**[0393]** In some embodiments, the reference electrode can be disposed remotely from the working electrode. In one embodiment, the reference electrode remains within the fluid flow, but is disposed within the fluid coupler. For example, the reference electrode can be coiled within the fluid coupler such that it is contact with saline flowing into the host, but it is not in physical contact with the host's blood (except when blood is withdrawn from the catheter). In another embodiment, the reference electrode is removed from fluid flow, but still maintains bodily fluid contact. For example, the reference electrode can be wired to an adhesive patch that is adhered to the host, such that the reference electrode is in contact with the host's skin. In yet another embodiment, the reference electrode can be external from the system, such as but not limited to in contact with the exterior of the *ex vivo* portion of the system, in fluid or electrical contact with a connected saline drip or other medical device, or in bodily contact, such as is generally done with EKG electrical contacts. While not wishing to be bound by theory, it is believed to locating the reference electrode remotely from the working electrode permits manufacture of a smaller sensor footprint (e.g., diameter) that will have relatively less affect on the host's blood flow, such as less thrombosis, than a sensor having a relatively larger footprint (e.g., wherein both the working electrode and the reference electrode are adjacent to each other and within the blood path).

**[0394]** In some embodiments of the sensor system, *in vivo* portion of the sensor (e.g., the tip 14a) has an enlarged area (e.g., a bulbous, nail head-shaped, football-shaped, cone-shaped, cylindrical, etc. portion) as compared a substantial portion of the sensor (e.g., diameter of the *in vivo* portion of the sensor). The sensor tip can be made bulbous by any convenient systems and methods known in the art, such as but not limited to arc welding, crimping, smashing, welding, molding, heating, and plasma arc welding. While not wishing to be bound by theory, it is believed that an enlarged sensor tip (e.g., bulbous) will prevent vessel piercing as the sensor is pushed forward into the vessel.

**[0395]** The sensor of the preferred embodiments is designed with a minimally invasive architecture so as to minimize reactions or effects on the blood flow (or on the sensor in the blood flow). Accordingly, the sensor designs described herein, consider minimization of dimensions and arrangement of the electrodes and other components of the sensor system, particularly the *in vivo* portion of the sensor (or any portion of the sensor in fluid contact with the blood flow).

**[0396]** Accordingly, in some embodiments, a substantial portion of the *in vivo* portion of the sensor is designed with at least one dimension less than about 0.020, 0.015, 0.012, 0.010, 0.008, 0.006, 0.005, 0.004 inches. In some embodiments, a substantial portion of the sensor that is in fluid contact with the blood flow is designed with at least one dimension less than about 0.015, 0.012, 0.010, 0.008, 0.006, 0.005, 0.004, 0.003, 0.002, 0.001 inches. As one exemplary embodiment, a sensor such as described in more detail with reference to Figs. 1A to 1C is formed from a 0.004 inch conductive wire (e.g., platinum) for a diameter of about 0.004 inches along a substantial portion of the sensor (e.g., *in vivo* portion or fluid contact portion). As another exemplary embodiment, a sensor such as described in more detail with reference to Figs. 1A to 1C is formed from a 0.004 inch conductive wire and vapor deposited with an insulator material for a diameter of about 0.005 inches along a substantial portion of the sensor (e.g., *in vivo* portion or fluid contact portion), after which a desired electroactive surface area can be exposed. In the above two exemplary embodiments, the reference electrode can be located remote from the working electrode (e.g., formed from the conductive wire). While the devices and methods described herein are directed to use within the host's blood stream, one skilled in the art will recognize that the systems, configurations, methods and principles of operation described herein can be incorporated into other analyte sensing devices, such as but not limited to subcutaneous devices or wholly implantable devices such as described in U.S. Patent Publication No. US-2006-0416700-A1.

**[0397]** **Fig. 3C** is a cross section of the sensor shown in **Fig. 3B,** taken at line C - C. Preferably, a membrane system (see **Fig. 3C**) is deposited over the electroactive surfaces of the sensor and includes a plurality of domains or layers, such as described in more detail below, with reference to **Figs. 3B** and **3C.** The membrane system can be deposited on the exposed electroactive surfaces using known thin film techniques (for example, spraying, electro-depositing, dipping, and the like). In one exemplary embodiment, each domain is deposited by dipping the sensor into a solution and drawing out the sensor at a speed that provides the appropriate domain thickness. In general, the membrane system can be disposed over (deposited on) the electroactive surfaces using methods appreciated by one skilled in the art.

**[0398]** In general, the membrane system includes a plurality of domains, for example, an electrode domain **347,** an interference domain **348,** an enzyme domain **349** (for example, including glucose oxidase), and a resistance domain **350,** as shown in Fig. 3C, and can include a high oxygen solubility domain, and/or a bioprotective domain (not shown),

such as is described in more detail in U.S. Patent Publication No. US-2005-0245799-A1, and such as is described in more detail below. The membrane system can be deposited on the exposed electroactive surfaces using known thin film techniques (for example, vapor deposition, spraying, electro-depositing, dipping, and the like). In alternative embodiments, however, other vapor deposition processes (e.g., physical and/or chemical vapor deposition processes) can be useful for providing one or more of the insulating and/or membrane layers, including ultrasonic vapor deposition, electrostatic deposition, evaporative deposition, deposition by sputtering, pulsed laser deposition, high velocity oxygen fuel deposition, thermal evaporator deposition, electron beam evaporator deposition, deposition by reactive sputtering molecular beam epitaxy, atmospheric pressure chemical vapor deposition (CVD), atomic layer CVD, hot wire CVD, low-pressure CVD, microwave plasma-assisted CVD, plasma-enhanced CVD, rapid thermal CVD, remote plasma-enhanced CVD, and ultra-high vacuum CVD, for example. However, the membrane system can be disposed over (or deposited on) the electroactive surfaces using any known method, as will be appreciated by one skilled in the art.

**[0399]** In some embodiments, one or more domains of the membrane systems are formed from materials such as described above in connection with the porous layer, such as silicone, polytetrafluoroethylene, polyethylene-co-tetrafluoroethylene, polyolefin, polyester, polycarbonate, biostable polytetrafluoroethylene, homopolymers, copolymers, terpolymers of polyurethanes, polypropylene (PP), polyvinylchloride (PVC), polyvinylidene fluoride (PVDF), polybutylene terephthalate (PBT), polymethylmethacrylate (PMMA), polyether ether ketone (PEEK), polyurethanes, cellulosic polymers, polysulfones and block copolymers thereof including, for example, di-block, tri-block, alternating, random and graft copolymers. U.S. Patent Publication No. US-2005-0245799-A1 describes biointerface and membrane system configurations and materials that may be applied to the preferred embodiments.

Electrode Domain

**[0400]** In selected embodiments, the membrane system comprises an electrode domain. The electrode domain **347** is provided to ensure that an electrochemical reaction occurs between the electroactive surfaces of the working electrode and the reference electrode, and thus the electrode domain **347** is preferably situated more proximal to the electroactive surfaces than the interference and/or enzyme domain. Preferably, the electrode domain includes a coating that maintains a layer of water at the electrochemically reactive surfaces of the sensor. In other words, the electrode domain is present to provide an environment between the surfaces of the working electrode and the reference electrode, which facilitates an electrochemical reaction between the electrodes. For example, a humectant in a binder material can be employed as an electrode domain; this allows for the full transport of ions in the aqueous environment. The electrode domain can also assist in stabilizing the operation of the sensor by accelerating electrode start-up and drifting problems caused by inadequate electrolyte. The material that forms the electrode domain can also provide an environment that protects against pH-mediated damage that can result from the formation of a large pH gradient due to the electrochemical activity of the electrodes.

**[0401]** In one embodiment, the electrode domain **347** includes a flexible, water-swellable, hydrogel film having a "dry film" thickness of from about 0.05 microns or less to about 20 microns or more, more preferably from about 0.05, 0.1, 0.15, 0.2, 0.25, 0.3, 0.35, 0.4, 0.45, 0.5, 1, 1.5, 2, 2.5, 3, or 3.5 microns to about 4, 5, 6, 7, 8, 9, 10, 11, 12, 13, 14, 15, 16, 17, 18, 19, or 19.5 microns, and more preferably still from about 3, 2.5, 2, or 1 microns, or less, to about 3.5, 4, 4.5, or 5 microns or more. "Dry film" thickness refers to the thickness of a cured film cast from a coating formulation by standard coating techniques.

**[0402]** In certain embodiments, the electrode domain **347** is formed of a curable mixture of a urethane polymer and a hydrophilic polymer. Particularly preferred coatings are formed of a polyurethane polymer having carboxylate or hydroxyl functional groups and nonionic hydrophilic polyether segments, wherein the polyurethane polymer is crosslinked with a water-soluble carbodiimide (*e.g.*, 1-ethyl-3-(3-dimethylaminopropyl)carbodiimide (EDC)) in the presence of polyvinylpyrrolidone and cured at a moderate temperature of about 50°C.

**[0403]** In some preferred embodiments, the electrode domain **347** is formed from a hydrophilic polymer (*e.g.*, a polyamide, a polylactone, a polyimide, a polylactam, a functionalized polyamide, a functionalized polylactone, a functionalized polyimide, a functionalized polylactam or a combination thereof) that renders the electrode domain substantially more hydrophilic than an overlying domain, (*e.g.*, interference domain, enzyme domain). In some embodiments, the electrode domain is formed substantially entirely and/or primarily from a hydrophilic polymer. In some embodiments, the electrode domain is formed substantially entirely from PVP. In some embodiments, the electrode domain is formed entirely from a hydrophilic polymer. Useful hydrophilic polymers include but are not limited to poly-N-vinylpyrrolidone (PVP), poly-N-vinyl-2-piperidone, poly-N-vinyl-2-caprolactam, poly-N-vinyl-3-methyl-2-caprolactam, poly-N-vinyl-3-methyl-2-piperidone, poly-N-vinyl-4-methyl-2-piperidone, poly-N-vinyl-4-methyl-2-caprolactam, poly-N-vinyl-3-ethyl-2-pyrrolidone, poly-N-vinyl-4,5-dimethyl-2-pyrrolidone, polyvinylimidazole, poly-N,N-dimethylacrylamide, polyvinyl alcohol, polyacrylic acid, polyethylene oxide, poly-2-ethyl-oxazoline, copolymers thereof and mixtures thereof. A blend of two or more hydrophilic polymers is preferred in some embodiments. In some preferred embodiments, the hydrophilic polymer(s) is not crosslinked. In alternative embodiments, crosslinking is preferred, such as by adding a crosslinking agent, such as but

not limited to EDC, or by irradiation at a wavelength sufficient to promote crosslinking between the hydrophilic polymer molecules, which is believed to create a more tortuous diffusion path through the domain.

**[0404]** An electrode domain formed from a hydrophilic polymer (e.g., PVP) has been shown to substantially reduce break-in time of analyte sensors; for example, a glucose sensor utilizing a cellulosic-based interference domain such as described in more detail elsewhere herein. In some embodiments, a uni-component electrode domain formed from a single hydrophilic polymer (e.g., PVP) has been shown to substantially reduce break-in time of a glucose sensor to less than about 2 hours, less than about 1 hour, less than about 20 minutes and/or substantially immediately, such as exemplified in Examples 9 through 11 and 13. Generally, sensor break-in is the amount of time required (after implantation) for the sensor signal to become substantially representative of the analyte concentration. Sensor break-in includes both membrane break-in and electrochemical break-in, which are described in more detail elsewhere herein. In some embodiments, break-in time is less than about 2 hours. In other embodiments, break-in time is less than about 1 hour. In still other embodiments, break-in time is less than about 30 minutes, less than about 20 minutes, less than about 15 minutes, less than about 10 minutes, or less. In a preferred embodiment, sensor break-in occurs substantially immediately. Advantageously, in embodiments wherein the break-in time is about 0 minutes (substantially immediately), the sensor can be inserted and begin providing substantially accurate analyte (*e.g.*, glucose) concentrations almost immediately post-insertion, for example, wherein membrane break-in does not limit start-up time.

**[0405]** While not wishing to be bound by theory, it is believed that providing an electrode domain that is substantially more hydrophilic than the next more distal membrane layer or domain (*e.g.*, the overlaying domain; the layer more distal to the electroactive surface than the electrode domain, such as an interference domain or an enzyme domain) reduces the break-in time of an implanted sensor, by increasing the rate at which the membrane system is hydrated by the surrounding host tissue. While not wishing to be bound by theory, it is believed that, in general, increasing the amount of hydrophilicity of the electrode domain relative to the overlaying layer (*e.g.*, the distal layer in contact with electrode domain, such as the interference domain, enzyme domain, etc.), increases the rate of water absorption, resulting in reduced sensor break-in time. The hydrophilicity of the electrode domain can be substantially increased by the proper selection of hydrophilic polymers, based on their hydrophilicity relative to each other and relative to the overlaying layer (*e.g.*, cellulosic-based interference domain), with preferred polymers being substantially more hydrophilic than the overlaying layer. In one exemplary embodiment, PVP forms the electrode domain, the interference domain is formed from a blend of cellulosic derivatives, such as but not limited to cellulose acetate butyrate and cellulose acetate; it is believed that since PVP is substantially more hydrophilic than the cellulosic- based interference domain, the PVP rapidly draws water into the membrane to the electrode domain, and enables the sensor to function with a desired sensitivity and accuracy and starting within a substantially reduced time period after implantation. Reductions in sensor break-in time reduce the amount of time a host must wait to obtain sensor readings, which is particularly advantageous not only in ambulatory applications, but particularly in hospital settings where time is critical.

**[0406]** While not wishing to be bound by theory, it is believed that when the water absorption of the overlying domain (*e.g.*, the domain overlying the electrode domain) is less than the water absorption of the electrode domain (*e.g.*, during membrane equilibration), then the difference in water absorption between the two domains will drive membrane equilibration and thus membrane break-in. Namely, increasing the difference in hydrophilicity (*e.g.*, between the two domains) results in an increase in the rate of water absorption, which, in turn, results in a decrease in membrane break-in time and/or sensor break-in time. As discussed elsewhere herein, the relative hydrophilicity of the electrode domain as compared to the overlying domain can be modulated by a selection of more hydrophilic materials for formation of the electrode domain (and/or more hydrophobic materials for the overlying domain(s)). For example, an electrode domain with hydrophilic polymer capable of absorbing larger amounts of water can be selected instead of a second hydrophilic polymer that is capable of absorbing less water than the first hydrophilic polymer. In some embodiments, the water content difference between the electrode domain and the overlying domain (*e.g.*, during or after membrane equilibration) is from about 1% or less to about 90% or more. In other embodiments, the water content difference between the electrode domain and the overlying domain is from about 10% or less to about 80% or more. In still other embodiments, the water content difference between the electrode domain and the overlying domain is from about 30% or less to about 60% or more. In preferred embodiments, the electrode domain absorbs 5 wt. % or less to 95 wt. % or more water, preferably 5, 10, 15, 20, 25, 30, 35, 40, 45, or 50 wt. % to about 55, 60, 65, 70, 75, 80, 85, 90 or 95 wt. % water than the adjacent (overlying) domain (*e.g.*, the domain that is more distal to the electroactive surface than the electrode domain).

**[0407]** In another example, the rate of water absorption by a polymer can be affected by other factors, such as but not limited to the polymer's molecular weight. For example, the rate of water absorption by PVP is dependent upon its molecular weight, which is typically from about 40 kDa or less to about 360 kDa or more; with a lower molecular weight PVP (*e.g.*, 40 kDa) absorbing water faster than a higher molecular weight PVP. Accordingly, modulating factors, such as molecular weight, that affect the rate of water absorption by a polymer, can promote the proper selection of materials for electrode domain fabrication. In one embodiment, a lower molecular weight PVP is selected, to reduce break-in time.

**[0408]** Preferably, the electrode domain is deposited by known thin film deposition techniques (*e.g.*, spray coating or dip-coating the electroactive surfaces of the sensor). In some embodiments, the electrode domain is formed by dip-

coating the electroactive surfaces in an electrode domain solution (*e.g.*, 5, 10, 15, 20, 25 or 30% or more PVP in deionized water) and curing the domain for a time of from about 15 minutes to about 30 minutes at a temperature of from about 40°C to about 55°C (and can be accomplished under vacuum (*e.g.*, 20 to 30 mmHg)). In embodiments wherein dip-coating is used to deposit the electrode domain, a preferred insertion rate of from about 1 to about 3 inches per minute into the electrode domain solution, with a preferred dwell time of from about 0.5 to about 2 minutes in the electrode domain solution, and a preferred withdrawal rate of from about 0.25 to about 2 inches per minute from the electrode domain solution provide a functional coating. However, values outside of those set forth above can be acceptable or even desirable in certain embodiments, for example, depending upon solution viscosity and solution surface tension, as is appreciated by one skilled in the art. In one embodiment, the electroactive surfaces of the electrode system are dip-coated one time (one layer) and cured at 50°C under vacuum for 20 minutes. In another embodiment, the electroactive surfaces of the electrode system is dip-coated and cured at 50°C under vacuum for 20 minutes a first time, followed by dip coating and curing at 50°C under vacuum for 20 minutes a second time (two layers). In still other embodiments, the electroactive surfaces can be dip-coated three or more times (three or more layers). In other embodiments, the 1, 2, 3 or more layers of PVP are applied to the electroactive surfaces by spray coating or vapor deposition. In some embodiments, a crosslinking agent (*e.g.*, EDC) can be added to the electrode domain casting solution to promote crosslinking within the domain (*e.g.*, between electrode domain polymer components, latex, etc.). In some alternative embodiments however, no crosslinking agent is used and the electrode domain is not substantially crosslinked.

**[0409]** In some embodiments, the deposited PVP electrode domain **347** has a "dry film" thickness of from about 0.05 microns or less to about 20 microns or more, more preferably from about 0.05, 0.1, 0.15, 0.2, 0.25, 0.3, 0.35, 0.4, 0.45, 0.5, 1, 1.5, 2, 2.5, 3, or 3.5 microns to about 4, 5, 6, 7, 8, 9, 10, 11, 12, 13, 14, 15, 16, 17, 18, 19, or 19.5 microns, and more preferably still from about 2, 2.5 or 3 microns to about 3.5, 4, 4.5, or 5 microns.

**[0410]** Although an independent electrode domain **347** is described herein, in some embodiments sufficient hydrophilicity can be provided in the interference domain and/or enzyme domain (the domain adjacent to the electroactive surfaces) so as to provide for the full transport of ions in the aqueous environment (*e.g.* without a distinct electrode domain). In these embodiments, an electrode domain is not necessary.

Interference Domain

**[0411]** Interferents are molecules or other species that are reduced or oxidized at the electrochemically reactive surfaces of the sensor, either directly or *via* an electron transfer agent, to produce a false positive analyte signal (*e.g.*, a non-analyte-related signal). This false positive signal causes the host's analyte concentration (*e.g.*, glucose concentration) to appear higher than the true analyte concentration. False-positive signal is a clinically significant problem in some conventional sensors. For example in a case of a dangerously hypoglycemic situation, wherein the host has ingested an interferent (*e.g.*, acetaminophen), the artificially high glucose signal can lead the host to believe that he is euglycemic (or, in some cases, hyperglycemic). As a result, the host can make inappropriate treatment decisions, such as taking no action, when the proper course of action is to begin eating. In another example, in the case of a euglycemic or hyperglycemic situation, wherein a host has consumed acetaminophen, an artificially high glucose signal caused by the acetaminophen can lead the host to believe that his glucose concentration is much higher than it truly is. Again, as a result of the artificially high glucose signal, the host can make inappropriate treatment decisions, such as giving himself too much insulin, which in turn can lead to a dangerous hypoglycemic episode.

**[0412]** In preferred embodiments, an interference domain **348** is provided that substantially restricts or blocks the flow of one or more interfering species therethrough; thereby substantially preventing artificial signal increases. Some known interfering species for a glucose sensor, as described in more detail herein, include acetaminophen, ascorbic acid, bilirubin, cholesterol, creatinine, dopamine, ephedrine, ibuprofen, L-dopa, methyl dopa, salicylate, tetracycline, tolazamide, tolbutamide, triglycerides, and uric acid. In general, the interference domain of the preferred embodiments is less permeable to one or more of the interfering species than to the measured species, *e.g.*, the product of an enzymatic reaction that is measured at the electroactive surface(s), such as but not limited to $H_2O_2$.

**[0413]** In one embodiment, the interference domain **348** is formed from one or more cellulosic derivatives. Cellulosic derivatives can include, but are not limited to, cellulose esters and cellulose ethers. In general, cellulosic derivatives include polymers such as cellulose acetate, cellulose acetate butyrate, 2-hydroxyethyl cellulose, cellulose acetate phthalate, cellulose acetate propionate, cellulose acetate trimellitate, and the like, as well as their copolymers and terpolymers with other cellulosic or non-cellulosic monomers. Cellulose is a polysaccharide polymer of β-D-glucose. While cellulosic derivatives are generally preferred, other polymeric polysaccharides having similar properties to cellulosic derivatives can also be employed in the preferred embodiments.

**[0414]** In one preferred embodiment, the interference domain **348** is formed from cellulose acetate butyrate. Cellulose acetate butyrate with a molecular weight of about 10,000 daltons to about 75,000 daltons, preferably from about 15,000, 20,000, or 25,000 daltons to about 50,000, 55,000, 60,000, 65,000, or 70,000 daltons, and more preferably about 20,000 daltons is employed. In certain embodiments, however, higher or lower molecular weights can be preferred. In some

embodiments, a blend of two or more cellulose acetate butyrates having different molecular weights is preferred. While a "blend" as defined herein (a composition of two or more substances that are not substantially chemically combined with each other and are capable of being separated) is generally preferred, in certain embodiments a single polymer incorporating different constituents (*e.g.*, separate constituents as monomeric units and/or substituents on a single polymer chain) can be employed instead. Additionally, a casting solution or dispersion of cellulose acetate butyrate at a wt. % of from about 5% to about 25%, preferably from about 5%, 6%, 7%, 8%, 9%, 10% 1%, 12%, 13%, 14% or 15% to about 16% 17%, 18%, 19%, 20%, 21%, 22%, 23%, 24% or 25%, and more preferably from about 5% to about 15% is preferred. Preferably, the casting solution includes a solvent or solvent system, for example an acetone:ethanol solvent system. Higher or lower concentrations can be preferred in certain embodiments. In alternative embodiments, a single solvent (e.g., acetone) is used to form a symmetrical membrane domain. A single solvent is used in casting solutions for forming symmetric membrane layer(s). A plurality of layers of cellulose acetate butyrate can be advantageously combined to form the interference domain in some embodiments, for example, three layers can be employed. It can be desirable to employ a mixture of cellulose acetate butyrate components with different molecular weights in a single solution, or to deposit multiple layers of cellulose acetate butyrate from different solutions comprising cellulose acetate butyrate of different molecular weights, different concentrations, and/or different chemistries (*e.g.*, functional groups). It can also be desirable to include additional substances in the casting solutions or dispersions, *e.g.*, functionalizing agents, crosslinking agents, other polymeric substances, substances capable of modifying the hydrophilicity/hydrophobicity of the resulting layer, and the like.

**[0415]** In one alternative embodiment, the interference domain **348** is formed from cellulose acetate. Cellulose acetate with a molecular weight of about 30,000 daltons or less to about 100,000 daltons or more, preferably from about 35,000, 40,000, or 45,000 daltons to about 55,000, 60,000, 65,000, 70,000, 75,000, 80,000, 85,000, 90,000, or 95,000 daltons, and more preferably about 50,000 daltons is preferred. In some embodiments, a blend of two or more cellulose acetates having different molecular weights is preferred. Additionally, a casting solution or dispersion of cellulose acetate at a weight percent of about 3% to about 10%, preferably from about 3.5%, 4.0%, 4.5%, 5.0%, 5.5%, 6.0%, or 6.5% to about 7.5%, 8.0%, 8.5%, 9.0%, or 9.5%, and more preferably about 8% is preferred. In certain embodiments, however, higher or lower molecular weights and/or cellulose acetate weight percentages can be preferred. It can be desirable to employ a mixture of cellulose acetates with molecular weights in a single solution, or to deposit multiple layers of cellulose acetate from different solutions comprising cellulose acetates of different molecular weights, different concentrations, or different chemistries (e.g., functional groups). It can also be desirable to include additional substances in the casting solutions or dispersions such as described in more detail above.

**[0416]** In addition to forming an interference domain from only cellulose acetate(s) or only cellulose acetate butyrate(s), the interference domain **348** can be formed from combinations or blends of cellulosic derivatives, such as but not limited to cellulose acetate and cellulose acetate butyrate, or combinations of layer(s) of cellulose acetate and layer(s) of cellulose acetate butyrate. In some embodiments, a blend of cellulosic derivatives (for formation of an interference domain) includes up to about 10 wt. % or more of cellulose acetate. For example, about 1, 2, 3, 4, 5, 6, 7, 8, 9 wt. % or more cellulose acetate is preferred, in some embodiments. In some embodiments, the cellulosic derivatives blend includes from about 90 wt. % or less to about 100 wt. % cellulose acetate butyrate. For example, in some embodiments, the blend includes about 91, 92, 93, 94, 95, 96, 97, 98 or 99 wt. % cellulose acetate butyrate. In some embodiments, the cellulosic derivative blend includes from about 1.5, 2.0, 2.5, 3.0 or 3.5 wt. % cellulose acetate to about 98.5, 98.0, 97.5, 97.0 or 96.5 wt. % cellulose acetate butyrate. In other embodiments, the blend includes from about 4, 4.5, 5, 5.5, 6, 6.5, 7, 7.5 or 8 wt. % cellulose acetate to about 96, 95.5, 95, 94.5, 94, 93.3, 93, 92.5 or 92 wt. % cellulose acetate butyrate. In still other embodiments, the blend includes from about 8.5, 9.0, 9.5, 10.0, 10.5 or 11.0 wt. % cellulose acetate to about 91.5, 91.0, 90.5, 90, 89.5 or 89 wt. % cellulose acetate butyrate.

**[0417]** In some embodiments, preferred blends of cellulose acetate and cellulose acetate butyrate contain from about 1.5 parts or less to about 60 parts or more cellulose acetate butyrate to one part of cellulose acetate. In some embodiments, a blend contains from about 2 parts to about 40 parts cellulose acetate butyrate to one part cellulose acetate. In other embodiments, about 4, 6, 8, 10, 12, 14, 16, 18 or 20 parts cellulose acetate butyrate to one part cellulose acetate is preferred for formation of the interference domain **348.** In still other embodiments, a blend having from 22, 24, 26, 28, 30, 32, 34, 36 or 38 parts cellulose acetate butyrate to one part cellulose acetate is preferred. As is discussed elsewhere herein, cellulose acetate butyrate is relatively more hydrophobic than cellulose acetate. Accordingly, the cellulose acetate/cellulose acetate butyrate blend contains substantially more hydrophobic than hydrophilic components.

**[0418]** Cellulose acetate butyrate is a cellulosic polymer having both acetyl and butyl groups, in addition to hydroxyl groups. Acetyl groups are more hydrophilic than butyl groups, and hydroxyl groups are more hydrophilic than both acetyl and butyl groups. Accordingly, the relative amounts of acetyl, butyl and hydroxyl groups can be used to modulate the hydrophilicity/hydrophobicity of the cellulose acetate butyrate of the cellulose acetate/cellulose acetate butyrate blend. A cellulose acetate butyrate can be selected based on the compound's relative amounts of acetate, butyrate and hydroxyl groups; and a cellulose acetate can be selected based on the compounds relative amounts of acetate and hydroxyl groups. For example, in some embodiments, a cellulose acetate butyrate having about 35 % or less acetyl groups, about

10 % to about 25 % butyl groups, and hydroxyl groups making up the remainder is preferred for formation of the interference domain **348.** In other embodiments a cellulose acetate butyrate having from about 25 % to about 34 % acetyl groups and from about 15 to about 20 % butyl groups is preferred. In still other embodiments, the preferred cellulose acetate butyrate contains from about 28 % to about 30 % acetyl groups and from about 16 to about 18 % butyl groups. In yet another embodiment, the cellulose acetate butyrate can have no acetate groups and from about 20 % to about 60 % butyrate groups. In yet another embodiment, the cellulose acetate butyrate has about 55 % butyrate groups and no acetate groups.

**[0419]** While an asymmetric interference domain can be used in some alternative embodiments, a symmetrical interference domain **348** (*e.g.*, of cellulosic-derivative blends, such as but not limited to blends of cellulose acetate components and cellulose acetate butyrate components) is preferred in some embodiments. Symmetrical membranes are uniform throughout their entire structure, without gradients of pore densities or sizes, or a skin on one side but not the other, for example. In various embodiments, a symmetrical interference domain **348** can be formed by the appropriate selection of a solvent (e.g., no anti-solvent is used), for making the casting solution. Appropriate solvents include solvents belonging to the ketone family that are able to solvate the cellulose acetate and cellulose acetate butyrate. The solvents include but are not limited to acetone, methyl ethyl ketone, methyl n-propyl ketone, cyclohexanone, and diacetone alcohol. Other solvents, such as furans (*e.g.*, tetra-hydro-furan and 1,4-dioxane), may be preferred in some embodiments. In one exemplary embodiment, from about 7 wt. % to about 9 wt. % solids (*e.g.*, a blend of cellulosic derivatives, such as cellulose acetate and cellulose acetate butyrate) are blended with a single solvent (*e.g.*, acetone), to form the casting solution for a symmetrical interference domain. In another embodiment, from about 10 to about 15% solids are blended with acetone to form the casting solution. In yet another embodiment, from about 16 to about 18% solids are blended with acetone to form the casting solution. A relatively lower or greater weight percent of solids is preferred to form the casting solution, in some embodiments.

**[0420]** The casting solution can be applied either directly to the electroactive surface(s) of the sensor or on top of an electrode domain layer (if included in the membrane system). The casting solution can be applied using any known thin film technique, as discussed elsewhere herein. Additionally, in various embodiments, a symmetrical interference domain **348** includes at least one layer; and in some embodiments, two, three or more layers are formed by the sequential application and curing of the casting solution.

**[0421]** The concentration of solids in the casting solution can be adjusted to deposit a sufficient amount of solids on the electrode in one layer (*e.g.*, in one dip or spray) to form a membrane layer with sufficient blocking ability, such that the equivalent glucose signal of an interferent (*e.g.*, compounds with an oxidation or reduction potential that overlaps with that of the measured species (*e.g.*, $H_2O_2$)), measured by the sensor, is about 60 mg/dL or less. For example, in some embodiments, the casting solution's percentage of solids is adjusted such that only a single layer (*e.g.*, dip one time) is required to deposit a sufficient amount of the cellulose acetate/cellulose acetate butyrate blend to form a functional symmetric interference domain that substantially blocks passage therethrough of at least one interferent, such as but not limited to acetaminophen, ascorbic acid, dopamine, ibuprofen, salicylic acid, tolbutamide, tetracycline, creatinine, uric acid, ephedrine, L-dopa, methyl dopa and tolazamide. In some embodiments, the amount of interference domain material deposited by as single dip is sufficient to reduce the equivalent glucose signal of the interferant (*e.g.*, measured by the sensor) to about 60 mg/dl or less. In preferred embodiments, the interferent's equivalent glucose signal response (measured by the sensor) is 50 mg/dl or less. In more preferred embodiments, the interferent produces an equivalent glucose signal response of 40 mg/dl or less. In still more preferred embodiments, the interferent produces an equivalent glucose signal response of less than about 30, 20 or 10 mg/dl. In one exemplary embodiment, the interference domain is configured to substantially block acetaminophen passage therethrough, wherein the equivalent glucose signal response of the acetaminophen is less than about 30 mg/dl.

**[0422]** In alternative embodiments, the interference domain **348** is configured to substantially block a therapeutic dose of acetaminophen. The term "therapeutic dose" as used herein is a broad term, and is to be given its ordinary and customary meaning to a person of ordinary skill in the art (and is not to be limited to a special or customized meaning), and refers without limitation to the quantity of any substance required to effect the cure of a disease, to relieve pain, or that will correct the manifestations of a deficiency of a particular factor in the diet, such as the effective dose used with therapeutically applied compounds, such as drugs. For example, a therapeutic dose of acetaminophen can be an amount of acetaminophen required to relieve headache pain or reduce a fever. As a further example, 1,000 mg of acetaminophen taken orally, such as by swallowing two 500 mg tablets of acetaminophen, is the therapeutic dose frequently taken for headaches. In some embodiments, the interference membrane is configured to block a therapeutic dose of acetaminophen, wherein the equivalent glucose signal response of the acetaminophen is less than about 60 mg/dl. In a preferred embodiment, the interference membrane is configured to block a therapeutic dose of acetaminophen, wherein the equivalent glucose signal response of the acetaminophen is less than about 40 mg/dl. In a more preferred embodiment, the interference membrane is configured to block a therapeutic dose of acetaminophen, wherein the equivalent glucose signal response of the acetaminophen is less than about 30 mg/dl.

**[0423]** While not wishing to be bound by theory, it is believed that, with respect to symmetrical cellulosic-based mem-

branes, there is an inversely proportional balance between interferent blocking and analyte sensitivity. Namely, changes to the interference domain configuration that increase interferent blocking can result in a corresponding decrease in sensor sensitivity. Sensor sensitivity is discussed in more detail elsewhere herein. It is believed that the balance between interferent blocking and sensor sensitivity is dependent upon the relative proportions of hydrophobic and hydrophilic components of the membrane layer (*e.g.*, the interference domain), with sensors having more hydrophobic interference domains having increased interferent blocking but reduces sensitivity; and sensors having more hydrophilic interference domains having reduced interferent blocking but increased sensitivity. It is believed that the hydrophobic and hydrophilic components of the interference domain can be balanced, to promote a desired level of interferent blocking while at the same time maintaining a desired level of analyte sensitivity. The interference domain hydrophobe-hydrophile balance can be manipulated and/or maintained by the proper selection and blending of the hydrophilic and hydrophobic interference domain components (*e.g.*, cellulosic derivatives having acetyl, butyryl, propionyl, methoxy, ethoxy, propoxy, hydroxyl, carboxymethyl, and/or carboxyethyl groups). For example, cellulose acetate is relatively more hydrophilic than cellulose acetate butyrate. In some embodiments, increasing the percentage of cellulose acetate (or reducing the percentage of cellulose acetate butyrate) can increase the hydrophilicity of the cellulose acetate/cellulose acetate butyrate blend, which promotes increased permeability to hydrophilic species, such as but not limited to glucose, $H_2O_2$ and some interferents (*e.g.*, acetaminophen). In another embodiment, the percentage of cellulose acetate butyrate is increased to increase blocking of interferants, but less permeability to some desired molecules, such as $H_2O_2$ and glucose, is also reduced.

**[0424]** One method, of manipulating the hydrophobe-hydrophile balance of the interference domain, is to select the appropriate percentages of acetyl groups (relatively more hydrophilic than butyl groups), butyl groups (relatively more hydrophobic than acetyl groups) and hydroxyl groups of the cellulose acetate butyrate used to form the interference domain **348.** For example, increasing the percentage of acetate groups on the cellulose acetate butyrate will make the cellulose acetate butyrate more hydrophilic. In another example, increasing the percentage of butyl groups on the cellulose acetate butyrate will make the cellulose acetate butyrate more hydrophobic. In yet another example, increasing the percentage of hydroxyl groups will increase the hydrophilicity of the cellulose acetate butyrate. Accordingly, the selection of a cellulose acetate butyrate that is more or less hydrophilic (or more or less hydrophobic) can modulate the over-all hydrophilicity of the cellulose acetate/cellulose acetate butyrate blend. In one exemplary embodiment, an interference domain can be configured to be relatively more hydrophobic (and therefore block interferants more strongly) by reducing the percentage of acetyl or hydroxyl groups or by increasing the percentage of butyl groups on the cellulose acetate butyrate used in the casting solution (while maintaining the relative ratio of cellulose acetate to cellulose acetate butyrate).

**[0425]** In some alternative embodiments, the interference domain **348** is formed of a blend of cellulosic derivatives, wherein the hydrophilic and hydrophobic components of the interference domain are balanced, such that the glucose sensitivity is from about 1 pA/mg/dL to about 100 pA/mg/dL, and at least one interferent is sufficiently blocked from passage through the interference domain such that the equivalent glucose signal response of the at least one interferent is less than about 60 mg/dL. In a preferred embodiment, the glucose sensitivity is from about 5 pA/mg/dL to about 25 pA/mg/dL. In a more preferred embodiments, the glucose sensitivity is from about 5 pA/mg/dL to about 25 pA/mg/dL and the equivalent glucose signal response of the at least one interferent is less than about 40 mg/dL. In a still more preferred embodiments, the glucose sensitivity is from about 5 pA/mg/dL to about 25 pA/mg/dL and the equivalent glucose signal response of the at least one interferent is less than about 30 mg/dL. In some embodiments, the balance between hydrophilic and hydrophobic components of the interference domain can be achieved by adjusting the amounts of hydrophilic and hydrophobic components, relative to each other, as well as adjusting the hydrophilic and hydrophobic groups (e.g., acetyl, butyryl, propionyl, methoxy, ethoxy, propoxy, hydroxyl, carboxymethyl, and/or carboxyethyl groups) of the components themselves (*e.g.*, cellulosic derivatives, such as but not limited to cellulose acetate and cellulose acetate butyrate).

**[0426]** In some alternative embodiments, additional polymers, such as Nafion®, can be used in combination with cellulosic derivatives to provide equivalent and/or enhanced function of the interference domain **348.** As one example, a layer of a 5 wt. % Nafion® casting solution was applied over a previously applied (*e.g.*, and cured) layer of 8 wt. % cellulose acetate, *e.g.,* by dip coating at least one layer of cellulose acetate and subsequently dip coating at least one layer Nafion® onto a needle-type sensor such as described with reference to the preferred embodiments. Any number of coatings or layers formed in any order may be suitable for forming the interference domain of the preferred embodiments.

**[0427]** In some alternative embodiments, more than one cellulosic derivative can be used to form the interference domain **348** of the preferred embodiments. In general, the formation of the interference domain on a surface utilizes a solvent or solvent system, in order to solvate the cellulosic derivative(s) (or other polymer) prior to film formation thereon. In preferred embodiments, acetone and ethanol are used as solvents for cellulose acetate; however one skilled in the art appreciates the numerous solvents that are suitable for use with cellulosic derivatives (and other polymers). Additionally, one skilled in the art appreciates that the preferred relative amounts of solvent can be dependent upon the

cellulosic derivative (or other polymer) used, its molecular weight, its method for deposition, its desired thickness, and the like. However, a percent solute of from about 1 wt. % to about 25 wt. % is preferably used to form the interference domain solution so as to yield an interference domain having the desired properties. The cellulosic derivative (or other polymer) used, its molecular weight, method for deposition, and desired thickness can be adjusted, depending upon one or more other of the parameters, and can be varied accordingly as is appreciated by one skilled in the art.

**[0428]** In some alternative embodiments, other polymer types that can be utilized as a base material for the interference domain **348** including polyurethanes, polymers having pendant ionic groups, and polymers having controlled pore size, for example. In one such alternative embodiment, the interference domain includes a thin, hydrophobic membrane that is non-swellable and restricts diffusion of high molecular weight species. The interference domain **48** is permeable to relatively low molecular weight substances, such as hydrogen peroxide, but restricts the passage of higher molecular weight substances, including glucose and ascorbic acid. Other systems and methods for reducing or eliminating interference species that can be applied to the membrane system of the preferred embodiments are described in U.S. Patent No. 7,074,307, U.S. Patent Publication No. US-2005-0176136-A1, U.S. Patent No. 7,081,195, and U.S. Patent Publication No. US-2005-0143635-A1. In some alternative embodiments, a distinct interference domain is not included.

**[0429]** In some embodiments, the interference domain **348** is deposited either directly onto the electroactive surfaces of the sensor or onto the distal surface of the electrode domain, for a domain thickness of from about 0.05 micron or less to about 20 microns or more, more preferably from about 0.05, 0.1, 0.15, 0.2, 0.25, 0.3, 0.35, 0.4, 0.45, 0.5, 1, 1.5, 2, 2.5, 3, or 3.5 microns to about 4, 5, 6, 7, 8, 9, 10, 11, 12, 13, 14, 15, 16, 17, 18, 19, or 19.5 microns, and more preferably still from about 1, 1.5 or 2 microns to about 2.5 or 3 microns. Thicker membranes can also be desirable in certain embodiments, but thinner membranes are generally preferred because they have a lower impact on the rate of diffusion of hydrogen peroxide from the enzyme membrane to the electrodes

**[0430]** In general, the membrane systems of the preferred embodiments can be formed and/or deposited on the exposed electroactive surfaces (*e.g.*, one or more of the working and reference electrodes) using known thin film techniques (for example, casting, spray coating, drawing down, electro-depositing, dip coating, and the like), however casting or other known application techniques can also be utilized. Preferably, the interference domain 348 is deposited by spray or dip coating. In one exemplary embodiment of a needle-type (transcutaneous) sensor such as described herein, the interference domain is formed by dip coating the sensor into an interference domain solution using an insertion rate of from about 0.5 inch/min to about 60 inches/min, preferably 1 inch/min, a dwell time of from about 0 minute to about 2 minutes, preferably about 1 minute, and a withdrawal rate of from about 0.5 inch/minute to about 60 inches/minute, preferably about 1 inch/minute, and curing (drying) the domain from about 1 minute to about 30 minutes, preferably from about 3 minutes to about 15 minutes (and can be accomplished at room temperature or under vacuum (*e.g.*, 20 to 30 mmHg)). In one exemplary embodiment including cellulose acetate butyrate interference domain, a 3-minute cure (*i.e.*, dry) time is preferred between each layer applied. In another exemplary embodiment employing a cellulose acetate interference domain, a 15 5 minute cure *(i.e.,* dry) time is preferred between each layer applied.

**[0431]** In some embodiments, the dip process can be repeated at least one time and up to 10 times or more. In other embodiments, only one dip is preferred. The preferred number of repeated dip processes depends upon the cellulosic derivative(s) used, their concentration, conditions during deposition (*e.g.*, dipping) and the desired thickness (*e.g.*, sufficient thickness to provide functional blocking of certain interferents), and the like. In some embodiments, 1 to 3 microns may be preferred for the interference domain thickness; however, values outside of these can be acceptable or even desirable in certain embodiments, for example, depending upon viscosity and surface tension, as is appreciated by one skilled in the art. In one exemplary embodiment, an interference domain is formed from three layers of cellulose acetate butyrate. In another exemplary embodiment, an interference domain is formed from 10 layers of cellulose acetate. In another embodiment, an interference domain is formed from 1 layer of a blend of cellulose acetate and cellulose acetate butyrate. In alternative embodiments, the interference domain can be formed using any known method and combination of cellulose acetate and cellulose acetate butyrate, as will be appreciated by one skilled in the art.

**[0432]** In some embodiments, the electroactive surface can be cleaned prior to application of the interference domain **348.** In some embodiments, the interference domain **348** of the preferred embodiments can be useful as a bioprotective or biocompatible domain, namely, a domain that interfaces with host tissue when implanted in an animal (*e.g.*, a human) due to its stability and biocompatibility.

Enzyme Domain

**[0433]** In preferred embodiments, the membrane system further includes an enzyme domain **349** disposed more distally from the electroactive surfaces than the interference domain **348;** however other configurations can be desirable. In the preferred embodiments, the enzyme domain provides an enzyme to catalyze the reaction of the analyte and its co-reactant, as described in more detail below. In the preferred embodiments of a glucose sensor, the enzyme domain includes glucose oxidase; however other oxidases, for example, galactose oxidase or uricase oxidase, can also be used.

**[0434]** For an enzyme-based electrochemical glucose sensor to perform well, the sensor's response is preferably

limited by neither enzyme activity nor co-reactant concentration. Because enzymes, including glucose oxidase, are subject to deactivation as a function of time even in ambient conditions, this behavior is compensated for in forming the enzyme domain. Preferably, the enzyme domain is constructed of aqueous dispersions of colloidal polyurethane polymers including the enzyme. However, in alternative embodiments the enzyme domain is constructed from an oxygen enhancing material, for example, silicone, or fluorocarbon, in order to provide a supply of excess oxygen during transient ischemia. Preferably, the enzyme is immobilized within the domain. See, e.g., U.S. Patent Publication No. US-2005-0054909-A1.

**[0435]** In preferred embodiments, the enzyme domain is deposited onto the interference domain for a domain thickness of from about 0.05 micron or less to about 20 microns or more, more preferably from about 0.05, 0.1, 0.15, 0.2, 0.25, 0.3, 0.35, 0.4, 0.45, 0.5, 1, 1.5, 2, 2.5, 3, or 3.5 microns to about 4, 5, 6, 7, 8, 9, 10, 11, 12, 13, 14, 15, 16, 17, 18, 19, or 19.5 microns, and more preferably still from about 2, 2.5 or 3 microns to about 3.5, 4, 4.5, or 5 microns. However in some embodiments, the enzyme domain can be deposited directly onto the electroactive surfaces. Preferably, the enzyme domain is deposited by spray or dip coating. In one embodiment of needle-type (transcutaneous) sensor such as described herein, the enzyme domain is formed by dip coating the interference domain coated sensor into an enzyme domain solution and curing the domain for from about 15 to about 30 minutes at a temperature of from about 40°C to about 55°C (and can be accomplished under vacuum (e.g., 20 to 30 mmHg)). In embodiments wherein dip coating is used to deposit the enzyme domain at room temperature, a preferred insertion rate of from about 0.25 inch per minute to about 3 inches per minute, with a preferred dwell time of from about 0.5 minutes to about 2 minutes, and a preferred withdrawal rate of from about 0.25 inch per minute to about 2 inches per minute provides a functional coating. However, values outside of those set forth above can be acceptable or even desirable in certain embodiments, for example, depending upon viscosity and surface tension, as is appreciated by one skilled in the art. In one embodiment, the enzyme domain is formed by dip coating two times (namely, forming two layers) in an enzyme domain solution and curing at 50°C under vacuum for 20 minutes. However, in some embodiments, the enzyme domain can be formed by dip coating and/or spray coating one or more layers at a predetermined concentration of the coating solution, insertion rate, dwell time, withdrawal rate, and/or desired thickness.

Resistance Domain

**[0436]** In preferred embodiments, the membrane system includes a resistance domain 350 disposed more distal from the electroactive surfaces than the enzyme domain. Although the following description is directed to a resistance domain for a glucose sensor, the resistance domain can be modified for other analytes and co-reactants as well.

**[0437]** There exists a molar excess of glucose relative to the amount of oxygen in blood; that is, for every free oxygen molecule in extracellular fluid, there are typically more than 100 glucose molecules present (see Updike et al., Diabetes Care 5:207-21(1982)). However, an immobilized enzyme-based glucose sensor employing oxygen as co-reactant is preferably supplied with oxygen in non-rate-limiting excess in order for the sensor to respond linearly to changes in glucose concentration, while not responding to changes in oxygen concentration. Specifically, when a glucose-monitoring reaction is oxygen limited, linearity is not achieved above minimal concentrations of glucose. Without a semipermeable membrane situated over the enzyme domain to control the flux of glucose and oxygen, a linear response to glucose levels can be obtained only for glucose concentrations of up to about 40 mg/dL. However, in a clinical setting, a linear response to glucose levels is desirable up to at least about 400 mg/dL.

**[0438]** The resistance domain includes a semipermeable membrane that controls the flux of oxygen and glucose to the underlying enzyme domain, preferably rendering oxygen in a non-rate-limiting excess. As a result, the upper limit of linearity of glucose measurement is extended to a much higher value than that which is achieved without the resistance domain. In one embodiment, the resistance domain exhibits an oxygen to glucose permeability ratio of from about 50:1 or less to about 400:1 or more, preferably about 200:1. As a result, one-dimensional reactant diffusion is adequate to provide excess oxygen at all reasonable glucose and oxygen concentrations found in the subcutaneous matrix (See Rhodes et al., Anal. Chem., 66:1520-1529 (1994)).

**[0439]** In alternative embodiments, a lower ratio of oxygen-to-glucose can be sufficient to provide excess oxygen by using a high oxygen solubility domain (for example, a silicone or fluorocarbon-based material or domain) to enhance the supply/transport of oxygen to the enzyme domain. If more oxygen is supplied to the enzyme, then more glucose can also be supplied to the enzyme without creating an oxygen rate-limiting excess. In alternative embodiments, the resistance domain is formed from a silicone composition, such as is described in U.S. Patent Publication No. US-2005-0090607-A1.

**[0440]** In a preferred embodiment, the resistance domain includes a polyurethane membrane with both hydrophilic and hydrophobic regions to control the diffusion of glucose and oxygen to an analyte sensor, the membrane being fabricated easily and reproducibly from commercially available materials. A suitable hydrophobic polymer component is a polyurethane, or polyetherurethaneurea. Polyurethane is a polymer produced by the condensation reaction of a diisocyanate and a difunctional hydroxyl-containing material. A polyurethaneurea is a polymer produced by the condensation reaction of a diisocyanate and a difunctional amine-containing material. Preferred diisocyanates include aliphatic diisocyanates containing from about 4 to about 8 methylene units. Diisocyanates containing cycloaliphatic moieties can

also be useful in the preparation of the polymer and copolymer components of the membranes of preferred embodiments. The material that forms the basis of the hydrophobic matrix of the resistance domain can be any of those known in the art as appropriate for use as membranes in sensor devices and as having sufficient permeability to allow relevant compounds to pass through it, for example, to allow an oxygen molecule to pass through the membrane from the sample under examination in order to reach the active enzyme or electrochemical electrodes. Examples of materials which can be used to make non-polyurethane type membranes include vinyl polymers, polyethers, polyesters, polyamides, inorganic polymers such as polysiloxanes and polycarbosiloxanes, natural polymers such as cellulosic and protein based materials, and mixtures or combinations thereof.

**[0441]** In a preferred embodiment, the hydrophilic polymer component is polyethylene oxide. For example, one useful hydrophobic-hydrophilic copolymer component is a polyurethane polymer that includes about 20% hydrophilic polyethylene oxide. The polyethylene oxide portions of the copolymer are thermodynamically driven to separate from the hydrophobic portions of the copolymer and the hydrophobic polymer component. The 20% polyethylene oxide-based soft segment portion of the copolymer used to form the final blend affects the water pick-up and subsequent glucose permeability of the membrane.

**[0442]** In some embodiments, the resistance domain is formed from a silicone polymer modified to allow analyte (e.g., glucose) transport.

**[0443]** In some embodiments, the resistance domain is formed from a silicone polymer/hydrophobic-hydrophilic polymer blend. In one embodiment, The hydrophobic-hydrophilic polymer for use in the blend may be any suitable hydrophobic-hydrophilic polymer, including but not limited to components such as polyvinylpyrrolidone (PVP), polyhydroxyethyl methacrylate, polyvinylalcohol, polyacrylic acid, polyethers such as polyethylene glycol or polypropylene oxide, and copolymers thereof, including, for example, di-block, tri-block, alternating, random, comb, star, dendritic, and graft copolymers (block copolymers are discussed in U.S. Patent Nos. 4,803,243 and 4,686,044, which are incorporated herein by reference). In one embodiment, the hydrophobic-hydrophilic polymer is a copolymer of poly(ethylene oxide) (PEO) and poly(propylene oxide) (PPO). Suitable such polymers include, but are not limited to, PEO-PPO diblock copolymers, PPO-PEO-PPO triblock copolymers, PEO-PPO-PEO triblock copolymers, alternating block copolymers of PEO-PPO, random copolymers of ethylene oxide and propylene oxide, and blends thereof. In some embodiments, the copolymers may be optionally substituted with hydroxy substituents. Commercially available examples of PEO and PPO copolymers include the PLURONIC® brand of polymers available from BASF®. In one embodiment, PLURONIC® F-127 is used. Other PLURONIC® polymers include PPO-PEO-PPO triblock copolymers (e.g., PLURONIC® R products). Other suitable commercial polymers include, but are not limited to, SYNPERONICS® products available from UNIQEMA®. U.S. Patent Publication No. US-2007-0244379-A1 describes systems and methods suitable for the resistance and/or other domains of the membrane system of the preferred embodiments.

**[0444]** In preferred embodiments, the resistance domain is deposited onto the enzyme domain to yield a domain thickness of from about 0.05 microns or less to about 20 microns or more, more preferably from about 0.05, 0.1, 0.15, 0.2, 0.25, 0.3, 0.35, 0.4, 0.45, 0.5, 1, 1.5, 2, 2.5, 3, or 3.5 microns to about 4, 5, 6, 7, 8, 9, 10, 11, 12, 13, 14, 15, 16, 17, 18, 19, or 19.5 microns, and more preferably still from about 2, 2.5 or 3 microns to about 3.5, 4, 4.5, or 5 microns. Preferably, the resistance domain is deposited onto the enzyme domain by vapor deposition, spray coating, or dip coating. In one preferred embodiment, spray coating is the preferred deposition technique. The spraying process atomizes and mists the solution, and therefore most or all of the solvent is evaporated prior to the coating material settling on the underlying domain, thereby minimizing contact of the solvent with the enzyme.

**[0445]** In another preferred embodiment, physical vapor deposition (e.g., ultrasonic vapor deposition) is used for coating one or more of the membrane domain(s) onto the electrodes, wherein the vapor deposition apparatus and process include an ultrasonic nozzle that produces a mist of micro-droplets in a vacuum chamber. In these embodiments, the micro-droplets move turbulently within the vacuum chamber, isotropically impacting and adhering to the surface of the substrate. Advantageously, vapor deposition as described above can be implemented to provide high production throughput of membrane deposition processes (e.g., at least about 20 to about 200 or more electrodes per chamber), greater consistency of the membrane on each sensor, and increased uniformity of sensor performance, for example, as described below.

**[0446]** In some embodiments, depositing the resistance domain (for example, as described in the preferred embodiments above) includes formation of a membrane system that substantially blocks or resists ascorbate (a known electrochemical interferant in hydrogen peroxide-measuring glucose sensors). While not wishing to be bound by theory, it is believed that during the process of depositing the resistance domain as described in the preferred embodiments, a structural morphology is formed that is characterized in that ascorbate does not substantially permeate therethrough.

**[0447]** In a preferred embodiment, the resistance domain is deposited on the enzyme domain by spray coating a solution of from about 1 wt. % to about 5 wt. % polymer and from about 95 wt. % to about 99 wt. % solvent. In spraying a solution of resistance domain material, including a solvent, onto the enzyme domain, it is desirable to mitigate or substantially reduce any contact with enzyme of any solvent in the spray solution that can deactivate the underlying enzyme of the enzyme domain. Tetrahydrofuran (THF) is one solvent that minimally or negligibly affects the enzyme of

the enzyme domain upon spraying. Other solvents can also be suitable for use, as is appreciated by one skilled in the art.

**[0448]** Although a variety of spraying or deposition techniques can be used, spraying the resistance domain material and rotating the sensor at least one time by 180° can typically provide adequate coverage by the resistance domain. Spraying the resistance domain material and rotating the sensor at least two times by 120° provides even greater coverage (one layer of 360° coverage), thereby ensuring resistivity to glucose, such as is described in more detail above.

**[0449]** In preferred embodiments, the resistance domain is spray coated and subsequently cured for a time of from about 15 minutes to about 90 minutes at a temperature of from about 40°C to about 60°C (and can be accomplished under vacuum (e.g., from 20 to 30 mmHg)). A cure time of up to about 90 minutes or more can be advantageous to ensure complete drying of the resistance domain.

**[0450]** In one embodiment, the resistance domain is formed by spray coating at least six layers (namely, rotating the sensor seventeen times by 120° for at least six layers of 360° coverage) and curing at 50°C under vacuum for 60 minutes. However, the resistance domain can be formed by dip coating or spray coating any layer or plurality of layers, depending upon the concentration of the solution, insertion rate, dwell time, withdrawal rate, and/or the desired thickness of the resulting film. Additionally, curing in a convention oven can also be employed.

**[0451]** In certain embodiments, a variable frequency microwave oven can be used to cure the membrane domains/layers. In general, microwave ovens directly excite the rotational mode of solvents. Consequently, microwave ovens cure coatings from the inside out rather than from the outside in as with conventional convection ovens. This direct rotational mode excitation is responsible for the typically observed "fast" curing within a microwave oven. In contrast to conventional microwave ovens, which rely upon a fixed frequency of emission that can cause arcing of dielectric (metallic) substrates if placed within a conventional microwave oven, Variable Frequency Microwave (VFM) ovens emit thousands of frequencies within 100 milliseconds, which substantially eliminates arcing of dielectric substrates. Consequently, the membrane domains/layers can be cured even after deposition on metallic electrodes as described herein. While not wishing to be bound by theory, it is believe that VFM curing can increase the rate and completeness of solvent evaporation from a liquid membrane solution applied to a sensor, as compared to the rate and completeness of solvent evaporation observed for curing in conventional convection ovens.

**[0452]** In certain embodiments, VFM is can be used together with convection oven curing to further accelerate cure time. In some sensor applications wherein the membrane is cured prior to application on the electrode (see, for example, U.S. Patent Publication No. US-2005-0245799-A1

**[0453]** ), conventional microwave ovens (e.g., fixed frequency microwave ovens) can be used to cure the membrane layer.

Treatment of Interference Domain/Membrane System

**[0454]** Although the above-described methods generally include a curing step in formation of the membrane system, including the interference domain, the preferred embodiments further include an additional treatment step, which can be performed directly after the formation of the interference domain and/or some time after the formation of the entire membrane system (or anytime in between). In some embodiments, the additional treatment step is performed during (or in combination with) sterilization of the sensor.

**[0455]** In some embodiments, the membrane system (or interference domain) is treated by exposure to ionizing radiation, for example, electron beam radiation, UV radiation, X-ray radiation, gamma radiation, and the like. Alternatively, the membrane can be exposed to visible light when suitable photoinitiators are incorporated into the interference domain. While not wishing to be bound by theory, it is believed that exposing the interference domain to ionizing radiation substantially crosslinks the interference domain and thereby creates a tighter, less permeable network than an interference domain that has not been exposed to ionizing radiation.

**[0456]** In some embodiments, the membrane system (or interference domain) is crosslinked by forming free radicals, which may include the use of ionizing radiation, thermal initiators, chemical initiators, photoinitiators (e.g., UV and visible light), and the like. Any suitable initiator or any suitable initiator system can be employed, for example, $\alpha$-hydroxyketone, $\alpha$-aminoketone, ammonium persulfate (APS), redox systems such as APS/bisulfite, or potassium permanganate. Suitable thermal initiators include but are not limited to potassium persulfate, ammonium persulfate, sodium persulfate, and mixtures thereof.

**[0457]** In embodiments wherein electron beam radiation is used to treat the membrane system (or interference domain), a preferred exposure time is from about 6k or 12kGy to about 25 or 50 kGy, more preferably about 25kGy. However, one skilled in the art appreciates that choice of molecular weight, composition of cellulosic derivative (or other polymer), and/or the thickness of the layer can affect the preferred exposure time of membrane to radiation. Preferably, the exposure is sufficient for substantially crosslinking the interference domain to form free radicals, but does not destroy or significantly break down the membrane or does not significantly damage the underlying electroactive surfaces.

**[0458]** In embodiments wherein UV radiation is employed to treat the membrane, UV rays from about 200 nm to about 400 nm are preferred; however values outside of this range can be employed in certain embodiments, dependent upon

the cellulosic derivative and/or other polymer used.

**[0459]** In some embodiments, for example, wherein photoinitiators are employed to crosslink the interference domain, one or more additional domains can be provided adjacent to the interference domain for preventing delamination that may be caused by the crosslinking treatment. These additional domains can be "tie layers" (*i.e.*, film layers that enhance adhesion of the interference domain to other domains of the membrane system). In one exemplary embodiment, a membrane system is formed that includes the following domains: resistance domain, enzyme domain, electrode domain, and cellulosic-based interference domain, wherein the electrode domain is configured to ensure adhesion between the enzyme domain and the interference domain. In embodiments wherein photoinitiators are employed to crosslink the interference domain, UV radiation of greater than about 290 nm is preferred. Additionally, from about 0.01 to about 1 wt % photoinitiator is preferred weight-to-weight with a preselected cellulosic polymer (*e.g.*, cellulose acetate); however values outside of this range can be desirable dependent upon the cellulosic polymer selected.

**[0460]** In general, sterilization of the transcutaneous sensor can be completed after final assembly, utilizing methods such as electron beam radiation, gamma radiation, glutaraldehyde treatment, and the like. The sensor can be sterilized prior to or after packaging. In an alternative embodiment, one or more sensors can be sterilized using variable frequency microwave chamber(s), which can increase the speed and reduce the cost of the sterilization process. In another alternative embodiment, one or more sensors can be sterilized using ethylene oxide (EtO) gas sterilization, for example, by treating with 100% ethylene oxide, which can be used when the sensor electronics are not detachably connected to the sensor and/or when the sensor electronics must undergo a sterilization process. In one embodiment, one or more packaged sets of transcutaneous sensors (e.g., 1, 2, 3, 4, or 5 sensors or more) are sterilized simultaneously.

Therapeutic Agents

**[0461]** A variety of therapeutic (bioactive) agents can be used with the analyte sensor system of the preferred embodiments, such as the analyte sensor system of the embodiments shown in **Figs. 1A - 3C.** In some embodiments, the therapeutic agent is an anticoagulant. The term "anticoagulant" as used herein is a broad term, and is to be given its ordinary and customary meaning to a person of ordinary skill in the art (and is not to be limited to a special or customized meaning), and refers without limitation to a substance the prevents coagulation (e.g., minimizes, reduces, or stops clotting of blood). In some embodiments, an anticoagulant is included in the analyte sensor system to prevent coagulation within or on the sensor (e.g., within or on the catheter or within or on the sensor). Suitable anticoagulants for incorporation into the sensor system include, but are not limited to, vitamin K antagonists (e.g., Acenocoumarol, Clorindione, Dicumarol (Dicoumarol), Diphenadione, Ethyl biscoumacetate, Phenprocoumon, Phenindione, Tioclomarol, or Warfarin), heparin group anticoagulants (e.g., Platelet aggregation inhibitors: Antithrombin III, Bemiparin, Dalteparin, Danaparoid, Enoxaparin, Heparin, Nadroparin, Pamaparin, Reviparin, Sulodexide, Tinzaparin), other platelet aggregation inhibitors (e.g., Abciximab, Acetylsalicylic acid (Aspirin), Aloxiprin, Beraprost, Ditazole, Carbasalate calcium, Cloricromen, Clopidogrel, Dipyridamole, Epoprostenol, Eptifibatide, Indobufen, Iloprost, Picotamide, Ticlopidine, Tirofiban, Treprostinil, Triflusal), enzymes (e.g., Alteplase, Ancrod, Anistreplase, Brinase, Drotrecogin alfa, Fibrinolysin, Protein C, Reteplase, Saruplase, Streptokinase, Tenecteplase, Urokinase), direct thrombin inhibitors (e.g., Argatroban, Bivalirudin, Desirudin, Lepirudin, Melagatran, Ximelagatran, other antithrombotics (e.g., Dabigatran, Defibrotide, Dermatan sulfate, Fondaparinux, Rivaroxaban) and the like.

**[0462]** In one embodiment, heparin is incorporated into the analyte sensor system. In a further embodiment, heparin is coated on the catheter (inner and/or outer diameter) and/or sensor, for example, by dipping or spraying. While not wishing to be bound by theory, it is believed that heparin coated on the catheter and/or sensor prevents aggregation and clotting of blood on the analyte sensor system, thereby preventing thromboembolization (e.g., prevention of blood flow by the thrombus or clot) and/or subsequent complications. In another embodiment, an antimicrobial is coated on the catheter (inner and/or outer diameter) and/or sensor.

**[0463]** In some embodiments, the therapeutic agent is an antimicrobial. The term "antimicrobial agent" as used in the preferred embodiments means antibiotics, antiseptics, disinfectants and synthetic moieties, and combinations thereof, that are soluble in organic solvents such as alcohols, ketones, ethers, aldehydes, acetonitrile, acetic acid, methylene chloride and chloroform.

**[0464]** Classes of antibiotics that can be used include tetracyclines (i.e. minocycline), rifamycins (i.e. rifampin), macrolides (i.e. erythromycin), penicillins (i.e. nafeillin), cephalosporins (i.e. cefazolin), other beta-lactam antibiotics (i.e. imipenem, aztreonam), aminoglycosides (i.e. gentamicin), chloramphenicol, sufonamides (i.e. sulfamethoxazole), glycopeptides (i.e. vancomycin), quinolones (i.e. ciprofloxacin), fusidic acid, trimethoprim, metronidazole, clindamycin, mupirocin, polyenes (i.e. amphotericin B), azoles (i.e. fluconazole) and beta-lactam inhibitors (i.e. sulbactam).

**[0465]** Examples of specific antibiotics that can be used include minocycline, rifampin, erythromycin, nafcillin, cefazolin, imipenem, aztreonam, gentamicin, sulfamethoxazole, vancomycin, ciprofloxacin, trimethoprim, metronidazole, clindamycin, teicoplanin, mupirocin, azithromycin, clarithromycin, ofloxacin, lomefloxacin, norfloxacin, nalidixic acid, sparfloxacin, pefloxacin, amifloxacin, enoxacin, fleroxacin, temafloxacin, tosufloxacin, clinafloxacin, sulbactam, clavulanic acid, am-

photericin B, fluconazole, itraconazole, ketoconazole, and nystatin.

**[0466]** Examples of antiseptics and disinfectants are hexachlorophene, cationic bisiguanides (i.e. chlorhexidine, cyclohexidine) iodine and iodophores (i.e. povidoneiodine), para-chloro-meta-xylenol, triclosan, furan medical preparations (i.e. nitrofurantoin, nitrofurazone), methenamine, aldehydes (glutaraldehyde, formaldehyde) and alcohols. Other examples of antiseptics and disinfectants will readily suggest themselves to those of ordinary skill in the art.

**[0467]** These antimicrobial agents can be used alone or in combination of two or more of them. The antimicrobial agents can be dispersed throughout the material of the sensor and/or catheter. The amount of each antimicrobial agent used to impregnate the medical device varies to some extent, but is at least of an effective concentration to inhibit the growth of bacterial and fungal organisms, such as staphylococci, gram-positive bacteria, gram-negative bacilli and Candida.

**[0468]** In some embodiments, the membrane system of the preferred embodiments preferably include a bioactive agent, which is incorporated into at least a portion of the membrane system, or which is incorporated into the device and adapted to diffuse through the membrane.

**[0469]** There are a variety of systems and methods by which the bioactive agent is incorporated into the membrane of the preferred embodiments. In some embodiments, the bioactive agent is incorporated at the time of manufacture of the membrane system. For example, the bioactive agent can be blended prior to curing the membrane system, or subsequent to membrane system manufacture, for example, by coating, imbibing, solvent-casting, or sorption of the bioactive agent into the membrane system. Although the bioactive agent is preferably incorporated into the membrane system, in some embodiments the bioactive agent can be administered concurrently with, prior to, or after insertion of the device intravascularly, for example, by oral administration, or locally, for example, by subcutaneous injection near the implantation site. A combination of bioactive agent incorporated in the membrane system and bioactive agent administration locally and/or systemically can be preferred in certain embodiments.

**[0470]** In general, a bioactive agent can be incorporated into the membrane system, and/or incorporated into the device and adapted to diffuse therefrom, in order to modify the tissue response of the host to the membrane. In some embodiments, the bioactive agent is incorporated only into a portion of the membrane system adjacent to the sensing region of the device, over the entire surface of the device except over the sensing region, or any combination thereof, which can be helpful in controlling different mechanisms and/or stages of thrombus formation. In some alternative embodiments however, the bioactive agent is incorporated into the device proximal to the membrane system, such that the bioactive agent diffuses through the membrane system to the host circulatory system.

**[0471]** The bioactive agent can include a carrier matrix, wherein the matrix includes one or more of collagen, a particulate matrix, a resorbable or non-resorbable matrix, a controlled-release matrix, and/or a gel. In some embodiments, the carrier matrix includes a reservoir, wherein a bioactive agent is encapsulated within a microcapsule. The carrier matrix can include a system in which a bioactive agent is physically entrapped within a polymer network. In some embodiments, the bioactive agent is cross-linked with the membrane system, while in others the bioactive agent is sorbed into the membrane system, for example, by adsorption, absorption, or imbibing. The bioactive agent can be deposited in or on the membrane system, for example, by coating, filling, or solvent casting. In certain embodiments, ionic and nonionic surfactants, detergents, micelles, emulsifiers, demulsifiers, stabilizers, aqueous and oleaginous carriers, solvents, preservatives, antioxidants, or buffering agents are used to incorporate the bioactive agent into the membrane system. The bioactive agent can be incorporated into a polymer using techniques such as described above, and the polymer can be used to form the membrane system, coatings on the membrane system, portions of the membrane system, and/or any portion of the sensor system.

**[0472]** The membrane system can be manufactured using techniques known in the art. The bioactive agent can be sorbed into the membrane system, for example, by soaking the membrane system for a length of time (for example, from about an hour or less to about a week or more, preferably from about 4, 8, 12, 16, or 20 hours to about 1, 2, 3, 4, 5, or 7 days).

**[0473]** The bioactive agent can be blended into uncured polymer prior to forming the membrane system. The membrane system is then cured and the bioactive agent thereby cross-linked and/or encapsulated within the polymer that forms the membrane system.

**[0474]** In yet another embodiment, microspheres are used to encapsulate the bioactive agent. The microspheres can be formed of biodegradable polymers, most preferably synthetic polymers or natural polymers such as proteins and polysaccharides. As used herein, the term polymer is used to refer to both to synthetic polymers and proteins. U.S. Patent 6,281,015, which is incorporated herein by reference in its entirety, discloses some systems and methods that can be used in conjunction with the preferred embodiments. In general, bioactive agents can be incorporated in (1) the polymer matrix forming the microspheres, (2) microparticle(s) surrounded by the polymer which forms the microspheres, (3) a polymer core within a protein microsphere, (4) a polymer coating around a polymer microsphere, (5) mixed in with microspheres aggregated into a larger form, or (6) a combination thereof. Bioactive agents can be incorporated as particulates or by co-dissolving the factors with the polymer. Stabilizers can be incorporated by addition of the stabilizers to the factor solution prior to formation of the microspheres.

**[0475]** The bioactive agent can be incorporated into a hydrogel and coated or otherwise deposited in or on the membrane system. Some hydrogels suitable for use in the preferred embodiments include cross-linked, hydrophilic, three-dimensional polymer networks that are highly permeable to the bioactive agent and are triggered to release the bioactive agent based on a stimulus.

**[0476]** The bioactive agent can be incorporated into the membrane system by solvent casting, wherein a solution including dissolved bioactive agent is disposed on the surface of the membrane system, after which the solvent is removed to form a coating on the membrane surface.

**[0477]** The bioactive agent can be compounded into a plug of material, which is placed within the device, such as is described in U.S. Patent Nos. 4,506,680 and 5,282,844.

**[0478]** In some embodiments, it is preferred to dispose the plug beneath a membrane system; in this way, the bioactive agent is controlled by diffusion through the membrane, which provides a mechanism for sustained-release of the bioactive agent in the host.

Release of Bioactive Agents

**[0479]** Numerous variables can affect the pharmacokinetics of bioactive agent release. The bioactive agents of the preferred embodiments can be optimized for short- and/or long-term release. In some embodiments, the bioactive agents of the preferred embodiments are designed to aid or overcome factors associated with short-term effects (e.g., acute inflammation and/or thrombosis) of sensor insertion. In some embodiments, the bioactive agents of the preferred embodiments are designed to aid or overcome factors associated with long-term effects, for example, chronic inflammation or build-up of fibrotic tissue and/or plaque material. In some embodiments, the bioactive agents of the preferred embodiments combine short- and long-term release to exploit the benefits of both.

**[0480]** As used herein, "controlled," "sustained," or "extended" release of the factors can be continuous or discontinuous, linear or non-linear. This can be accomplished using one or more types of polymer compositions, drug loadings, selections of excipients or degradation enhancers, or other modifications, administered alone, in combination or sequentially to produce the desired effect.

**[0481]** Short-term release of the bioactive agent in the preferred embodiments generally refers to release over a period of from about a few minutes or hours to about 2, 3, 4, 5, 6, or 7 days or more.

Loading of Bioactive Agents

**[0482]** The amount of loading of the bioactive agent into the membrane system can depend upon several factors. For example, the bioactive agent dosage and duration can vary with the intended use of the membrane system, for example, the intended length of use of the device and the like; differences among patients in the effective dose of bioactive agent; location and methods of loading the bioactive agent; and release rates associated with bioactive agents and optionally their carrier matrix. Therefore, one skilled in the art will appreciate the variability in the levels of loading the bioactive agent, for the reasons described above.

**[0483]** In some embodiments, wherein the bioactive agent is incorporated into the membrane system without a carrier matrix, the preferred level of loading of the bioactive agent into the membrane system can vary depending upon the nature of the bioactive agent. The level of loading of the bioactive agent is preferably sufficiently high such that a biological effect (e.g., thrombosis prevention) is observed. Above this threshold, bioactive agent can be loaded into the membrane system so as to imbibe up to 100% of the solid portions, cover all accessible surfaces of the membrane, and/or fill up to 100% of the accessible cavity space. Typically, the level of loading (based on the weight of bioactive agent(s), membrane system, and other substances present) is from about 1 ppm or less to about 1000 ppm or more, preferably from about 2, 3, 4, or 5 ppm up to about 10, 25, 50, 75, 100, 200, 300, 400, 500, 600, 700, 800, or 900 ppm. In certain embodiments, the level of loading can be 1 wt. % or less up to about 50 wt. % or more, preferably from about 2, 3, 4, 5, 6, 7, 8, 9, 10, 15, or 20 wt. % up to about 25, 30, 35, 40, or 45 wt. %.

**[0484]** When the bioactive agent is incorporated into the membrane system with a carrier matrix, such as a gel, the gel concentration can be optimized, for example, loaded with one or more test loadings of the bioactive agent. It is generally preferred that the gel contain from about 0.1 or less to about 50 wt. % or more of the bioactive agent(s), preferably from about 0.2, 0.3, 0.4, 0.5, 0.6, 0.7, 0.8, or 0.9 wt. % to about 6, 7, 8, 9, 10, 15, 20, 25, 30, 35, 40, or 45 wt. % or more bioactive agent(s), more preferably from about 1, 2, or 3 wt. % to about 4 or 5 wt. % of the bioactive agent(s). Substances that are not bioactive can also be incorporated into the matrix.

**[0485]** Referring now to microencapsulated bioactive agents, the release of the agents from these polymeric systems generally occurs by two different mechanisms. The bioactive agent can be released by diffusion through aqueous filled channels generated in the dosage form by the dissolution of the agent or by voids created by the removal of the polymer solvent or a pore forming agent during the original micro-encapsulation. Alternatively, release can be enhanced due to the degradation of the encapsulating polymer. With time, the polymer erodes and generates increased porosity and

microstructure within the device. This creates additional pathways for release of the bioactive agent.

**[0486]** In some embodiments, the sensor is designed to be bioinert, e.g., by the use of bioinert materials. Bioinert materials do not substantially cause any response from the host. As a result, cells can live adjacent to the material but do not form a bond with it. Bioinert materials include but are not limited to alumina, zirconia, titanium oxide or other bioinert materials generally used in the "catheter/catheterization" art. While not wishing to be bound by theory, it is believed that inclusion of a bioinert material in or on the sensor can reduce attachment of blood cells or proteins to the sensor, thrombosis or other host reactions to the sensor.

Dual-Electrode Analyte Sensors

**[0487]** In general, electrochemical analyte sensors provide at least one working electrode and at least one reference electrode, which are configured to generate a signal associated with a concentration of the analyte in the host, such as described herein, and as appreciated by one skilled in the art. The output signal is typically a raw data stream that is used to provide a useful value of the measured analyte concentration in a host to the patient or doctor, for example. However, the analyte sensors of the preferred embodiments may further measure at least one additional signal. For example, in some embodiments, the additional signal is associated with the baseline and/or sensitivity of the analyte sensor, thereby enabling monitoring of baseline and/or sensitivity changes that may occur in a continuous analyte sensor over time.

**[0488]** In preferred embodiments, the analyte sensor comprises a first working electrode **E1** and a second working electrode **E2**, in addition to a reference electrode, which is referred to as a dual-electrode system herein. The first and second working electrodes may be in any useful conformation, as described in US Patent Publications Nos. US-2007-0027385-A1, US-2007-0213611-A1, US-2007-0027284-A1, US-2007-0032717-A1, US-2007-0093704, and US Patent Application No. 11/865,572 filed on October 1, 2007 and entitled "DUAL-ELECTRODE SYSTEM FOR A CONTINUOUS ANALYTE SENSOR,".

**[0489]** In some preferred embodiments, the first and second working electrodes are twisted and/or bundled. For example, two wire working electrodes can be twisted together, such as in a helix conformation. The reference electrode can then be wrapped around the twisted pair of working electrodes. In some preferred embodiments, the first and second working electrodes include a coaxial configuration. A variety of dual-electrode system configurations are described with reference to **Figs. 7A1** through **11** of the references incorporated above. In some embodiments, the sensor is configured as a dual electrode sensor, such as described in US Patent Publication Nos. US-2005-0143635-A1; US-2007-0027385-A1; and US-2007-0213611-A1, and co-pending U.S. Patent Application No. 11/865,572.

**[0490]** However, a dual-electrode system can be provided in any planar or non-planar configuration, such as can be appreciated by one skilled in the art, and can be found in U.S. Patent 6,175,752 to Say et al.; U.S. Patent 6,579,690 to Bonnecaze et al.; U.S. Patent 6,484,046 to Say et al.; U.S. Patent 6,512,939 to Colvin et al.; U.S. Patent 6,477,395 to Schulman et al.; U.S. Patent 6,424,847 to Mastrototaro et al.; U.S. Patent 6,212,416 to Ward et al.; U.S. Patent 6,119,028 to Schulman et al.; U.S. Patent 6,400,974 to Lesho; U.S. Patent 6,595,919 to Berner et al.; U.S. Patent 6,141,573 to Kurnik et al.; 6,122,536 to Sun et al.; European Patent Application EP 1153571 to Varall et al.; U.S. Patent 6,512,939 to Colvin et al.; U.S. Patent 5,605,152 to Slate et al.; U.S. Patent 4,431,004 to Bessman et al.; U.S. Patent 4,703,756 to Gough et al.; U.S. Patent 6,514,718 to Heller et al.; U.S. Patent to 5,985,129 to Gough et al.; WO Patent Application Publication No. 04/021877 to Caduff; U.S. Patent 5,494,562 to Maley et al.; U.S. Patent 6,120,676 to Heller et al.; and U.S. Patent 6,542,765 to Guy et al.

**[0491]** In general, it is understood that the disclosed embodiments are applicable to a variety of continuous analyte measuring device configurations

**[0492]** **Fig. 3D** illustrates a dual-electrode system in preferred embodiments. The dual-electrode sensor system includes a first working electrode **E1** and the second working electrode **E2,** both of which are disposed beneath a sensor membrane **M02,** such as but not limited to a membrane system similar to that described with reference to **Fig. 3C** and/or **Figs. 3F** through **3I**. The first working electrode **E1** is disposed beneath an active enzymatic portion **M04** of the sensor membrane **M02**, which includes an enzyme configured to detect the analyte or an analyte-related compound. Accordingly, the first working electrode **E1** is configured to generate a first signal composed of both signal related to the analyte and signal related to non-analyte electroactive compounds (e.g., physiological baseline, interferents, and non-constant noise) that have an oxidation/reduction potential that overlaps with the oxidation/reduction potential of the analyte. This oxidation/reduction potential may be referred to as a "first oxidation/reduction potential" herein. The second working electrode **E2** is disposed beneath an inactive-enzymatic or non-enzymatic portion **M06** of the sensor membrane **M02**. The non-enzymatic portion **M06** of the membrane includes either an inactivated form of the enzyme contained in the enzymatic portion **M04** of the membrane or no enzyme. In some embodiments, the non-enzymatic portion **M06** can include a nonspecific protein, such as BSA, ovalbumin, milk protein, certain polypeptides, and the like. The non-enzymatic portion **M06** generates a second signal associated with noise of the analyte sensor: The noise of the sensor comprises signal contribution due to non-analyte electroactive species (e.g., interferents) that have an oxidation/reduction potential that

substantially overlaps the first oxidation/reduction potential (e.g., that overlap with the oxidation/reduction potential of the analyte). In some embodiments of a dual-electrode analyte sensor configured for fluid communication with a host's circulatory system, the non-analyte related electroactive species comprises at least one species selected from the group consisting of interfering species, non-reaction-related hydrogen peroxide, and other electroactive species.

**[0493]** In one exemplary embodiment, the dual-electrode analyte sensor is a glucose sensor having a first working electrode **E1** configured to generate a first signal associated with both glucose and non-glucose related electroactive compounds that have a first oxidation/reduction potential. Non-glucose related electroactive compounds can be any compound, in the sensor's local environment that has an oxidation/reduction potential substantially overlapping with the oxidation/reduction potential of $H_2O_2$, for example. While not wishing to be bound by theory, it is believed that the glucose-measuring electrode can measure both the signal directly related to the reaction of glucose with GOx (produces $H_2O_2$ that is oxidized at the working electrode) and signals from unknown compounds that are in the blood surrounding the sensor. These unknown compounds can be constant or non-constant (e.g., intermittent or transient) in concentration and/or effect. In some circumstances, it is believed that some of these unknown compounds are related to the host's disease state. For example, it is known that blood chemistry changes dramatically during/after a heart attack (e.g., pH changes, changes in the concentration of various blood components/protein, and the like). Additionally, a variety of medicaments or infusion fluid components (e.g., acetaminophen, ascorbic acid, dopamine, ibuprofen, salicylic acid, tolbutamide, tetracycline, creatinine, uric acid, ephedrine, L-dopa, methyl dopa and tolazamide) that may be given to the host may have oxidation/reduction potentials that overlap with that of $H_2O_2$.

**[0494]** In this exemplary embodiment, the dual-electrode analyte sensor includes a second working electrode **E2** that is configured to generate a second signal associated with the non-glucose related electroactive compounds that have the same oxidation/reduction potential as the above-described first working electrode (e.g., *para supra*). In some embodiments, the non-glucose related electroactive species includes at least one of interfering species, non-reaction-related $H_2O_2$, and other electroactive species. For example, interfering species includes any compound that is not directly related to the electrochemical signal generated by the glucose-GOx reaction, such as but not limited to electroactive species in the local environment produces by other bodily processes (e.g., cellular metabolism, a disease process, and the like). Other electroactive species includes any compound that has an oxidation/reduction potential similar to or overlapping that of $H_2O_2$.

**[0495]** The non-analyte (e.g., non-glucose) signal produced by compounds other than the analyte (e.g., glucose) obscured the signal related to the analyte, contributes to sensor inaccuracy, and is considered background noise. As described in greater detail in the section entitled "Noise Reduction," background noise includes both constant and non-constant components and must be removed to accurately calculate the analyte concentration. While not wishing to be bound by theory, it is believed that the sensor of the preferred embodiments are designed (e.g., with symmetry, coaxial design and/or integral formation, and interference domain of the membrane described elsewhere herein) such that the first and second electrodes are influenced by substantially the same external/environmental factors, which enables substantially equivalent measurement of both the constant and non-constant species/noise. This advantageously allows the substantial elimination of noise on the sensor signal (using electronics described elsewhere herein) to substantially reduce or eliminate signal effects due to noise, including non-constant noise (e.g., unpredictable biological, biochemical species, medicaments, pH fluctuations, $O_2$ fluctuations, or the like) known to effect the accuracy of conventional continuous sensor signals. Preferably, the sensor includes electronics operably connected to the first and second working electrodes. The electronics are configured to provide the first and second signals that are used to generate glucose concentration data substantially without signal contribution due to non-glucose-related noise. Preferably, the electronics include at least a potentiostat that provides a bias to the electrodes. In some embodiments, sensor electronics are configured to measure the current (or voltage) to provide the first and second signals. The first and second signals are used to determine the glucose concentration substantially without signal contribution due to non-glucose-related noise such as by but not limited to subtraction of the second signal from the first signal or alternative data analysis techniques. In some embodiments, the sensor electronics include a transmitter that transmits the first and second signals to a receiver, where additional data analysis and/or calibration of glucose concentration can be processed. U.S. Patent Publication No. US-2005-0027463-A1, US-2005-0203360-A1 and U.S. Patent Publication No. US-2006-0036142-A1 describe systems and methods for processing sensor analyte data.

**[0496]** In preferred embodiments, the dual-electrode sensor is configured such that the first and second working electrodes **E1, E2** are equivalently influenced by *in vivo* environmental factors. For example, in one embodiment, the dual-electrode sensor is configured for fluid communication with the circulatory system of the host, such as by implantation in the host's vein or artery *via* a vascular access device (also referred to as a fluid communication device herein) such as a catheter and/or cannula. When the sensor is contacted with a sample of the host's circulatory system (e.g., blood), the first and second working electrodes **E1**, **E2** are configured such that they are equivalently influenced by a variety of environmental factors impinging upon the sensor, such as but not limited to non-analyte related electroactive species (e.g., interfering species, non-reaction-related $H_2O_2$, an other electroactive species). Because the first and second working electrodes are equivalently influenced by *in vivo* environmental factors, the signal component associated with

the *in* vivo environmental factors (e.g., non-analyte related species with an oxidation/reduction potential that overlaps with that of the analyte) can be removed from the signal detected by the first working electrode (e.g., the first signal). This can give a substantially analyte-only signal. The effects of *in vivo* environmental factors upon the dual-electrode system are discussed in greater detail elsewhere herein with reference to **Figs. 3G-3I.**

**[0497]** In preferred embodiments, the dual-electrode sensor includes electronics (e.g., a processor module, processing memory) that are operably connected to the first and second working electrodes and are configured to provide the first and second signals to generate analyte concentration data substantially without signal contribution due to non-analyte-related noise. For example, the sensor electronics process and/or analyze the signals from the first and second working electrodes and calculate the portion of the first electrode signal that is due to analyte concentration only. The portion of the first electrode signal that is not due to the analyte concentration can be considered to be background, such as but not limited to noise. Accordingly, in one embodiment of a dual-electrode sensor system configured for fluid communication with a host's circulatory system (e.g., *via* a vascular access device) the system comprising electronics operably connected to the first and second working electrodes; the electronics are configured to process the first and second signals to generate analyte concentration data substantially without signal contribution due to noise.

**[0498]** As a non-limiting example, **Fig. 3E** illustrates one preferred embodiment, the dual-electrode analyte sensor. In this embodiment, the sensor comprises a first working electrode **E1** configured to detect the analyte and a second working electrode **E2**, wherein the first and second working electrodes are formed of two wire working electrodes twisted together to form a "twisted pair." The first working electrode **E1** is disposed beneath an enzymatic portion of the membrane (not shown) containing an analyte-detecting enzyme. For example, in a glucose-detecting dual-electrode analyte sensor, a glucose-detecting enzyme, such as GOX, is included in the enzymatic portion of the membrane. Accordingly, the first working electrode **E1** detects signal due to both the analyte and non-analyte-related species that have an oxidation/reduction potential that substantially overlaps with the oxidation/reduction potential of the analyte. The second working electrode **E2** is disposed beneath a portion of the membrane comprising either inactivated enzyme (e.g., inactivated by heat, chemical or UV treatment) or no enzyme. Accordingly, the second working electrode **E2** detects a signal associated with only the non-analyte electroactive species that have an oxidation/reduction potential that substantially overlaps with that of analyte. For example, in the glucose-detecting dual-electrode analyte sensor described above, the non-analyte (e.g., non-glucose) electroactive species have an oxidation/reduction potential that overlaps substantially with that of $H_2O_2$. A reference electrode **R**, such as a silver/silver chloride wire electrode, is wrapped around the twisted pair. The three electrodes **E1**, **E2** and **R** are connected to sensor electronics (not shown), such as described elsewhere herein. In preferred embodiments, the dual-electrode sensor is configured to provide an analyte-only signal (e.g., glucose-only signal) substantially without a signal component due to the non-analyte electroactive species (e.g., noise). For example, the dual-electrode sensor is operably connected to sensor electronics that process the first and second signals, such that a substantially analyte-only signal is provided (e.g., output to a user). In other exemplary embodiments, the dual-electrode sensor can be configured for detection of a variety of analytes other than glucose, such as but not limited to urea, creatinine, succinate, glutamine, oxygen, electrolytes, cholesterol, lipids, triglycerides, hormones, liver enzymes, and the like.

**[0499]** With reference to the analyte sensor embodiments disclosed herein, the surface area of the electroactive portion of the reference (and/or counter) electrode is at least six times the surface area of the working electrodes. In other embodiments, the reference (and/or counter) electrode surface is 1, 2, 3, 4, 5, 7, 8, 9 or 10 times the surface area of the working electrodes. In other embodiments, the reference (and/or counter) electrode surface area is 11, 12, 13, 14, 15, 16, 17, 18, 19 or 20 times the surface area of the working electrodes. For example, in a needle-type glucose sensor, similar to the embodiment shown in **Fig. 3E,** the surface area of the reference electrode (e.g., **R**) includes the exposed surface of the reference electrode, such as but not limited to the electrode surface facing away from the working electrodes **E1, E2.**

**[0500]** In various embodiments, the electrodes can be stacked or grouped similar to that of a leaf spring configuration, wherein layers of electrode and insulator (or individual insulated electrodes) are stacked in offset layers. The offset layers can be held together with bindings of non-conductive material, foil, or wire. As is appreciated by one skilled in the art, the strength, flexibility, and/or other material property of the leaf spring-configured or stacked sensor can be either modified (e.g., increased or decreased), by varying the amount of offset, the amount of binding, thickness of the layers, and/or materials selected and their thicknesses, for example.

**[0501]** In preferred embodiments, the analyte sensor substantially continuously measures the host's analyte concentration. In some embodiments, for example, the sensor can measure the analyte concentration every fraction of a second, about every fraction of a minute or every minute. In other exemplary embodiments, the sensor measures the analyte concentration about every 2, 3, 4, 5, 6, 7, 8, 9, or 10 minutes. In still other embodiments, the sensor measures the analyte concentration every fraction of an hour, such as but not limited to every 15, 30 or 45 minutes. Yet in other embodiments, the sensor measures the analyte concentration about every hour or longer. In some exemplary embodiments, the sensor measures the analyte concentration intermittently or periodically. In one preferred embodiment, the analyte sensor is a glucose sensor and measures the host's glucose concentration about every 4-6 minutes. In a further embodiment, the

sensor measures the host's glucose concentration every 5 minutes.

**[0502]** As a non-limiting example, dual-electrode glucose sensor can be manufactured as follows. In one embodiment, the working electrodes are first coated with a layer of insulating material (e.g., non-conductive material or dielectric) to prevent direct contact between the working electrodes **E1, E2** and the reference electrode **R.** At this point, or at any point hereafter, the two working electrodes can be twisted and/or bundled to form a twisted pair. A portion of the insulator on an exterior surface of each working electrode is etched away, to expose the electrode's electroactive surface. In some embodiments, an enzyme solution (e.g., containing active GOx) is applied to the electroactive surfaces of both working electrodes, and dried. Thereafter, the enzyme applied to one of the electroactive surfaces is inactivated. As is known in the art, enzymes can be inactivated by a variety of means, such as by heat, treatment with inactivating (e.g., denaturing) solvents, proteolysis, laser irradiation or UV irradiation (e.g., at 254-320 nm). For example, the enzyme coating one of the electroactive surfaces can be inactivated by masking one of the electroactive surfaces/electrodes (e.g., **E1,** temporarily covered with a UV-blocking material); irradiating the sensor with UV light (e.g., 254-320 nm; a wavelength that inactivates the enzyme, such as by cross-linking amino acid residues) and removing the mask. Accordingly, the GOx on **E2** is inactivated by the UV treatment, but the **E1** GOx is still active due to the protective mask. In other embodiments, an enzyme solution containing active enzyme is applied to a first electroactive surface (e.g., **E1**) and an enzyme solution containing either inactivated enzyme or no enzyme is applied to the second electroactive surface (e.g., **E2**). Thus, the enzyme-coated first electroactive surface (e.g., **E1**) detects analyte-related signal and non-analyte-related signal; while the second electroactive surface (e.g., **E2**), which lacks active enzyme, detects non-analyte-related signal. As described herein, the sensor electronics can use the data collected from the two working electrodes to calculate the analyte-only signal.

**[0503]** In some circumstances, cross talk can interfere with analyte/noise detection. In general, cross talk occurs when signal (e.g., in the form of energy and/or a detectable species such as but not limited to $H_2O_2$) is transferred from one electrode (e.g., the first working electrode) to another (e.g., the second working electrode), and detected as a signal by the other electrode. To prevent cross talk, in preferred embodiments, the first and second working electrodes **E1, E2** are separated by diffusion barrier, such as an insulator, a non-conductive material, a reference electrode and/or the like.

**[0504]** **Fig. 3F** illustrates the use of a diffusion barrier to prevent cross talk in a dual-electrode glucose sensor, in one embodiment. The first and second working electrodes **E1, E2** are disposed beneath a membrane **348** and separated by a diffusion barrier ***D.*** Within the membrane, glucose is metabolized by the GOx enzyme, which produces $H_2O_2$. The $H_2O_2$ produced by the enzymatic reaction can diffuse in any direction through the membrane **348.** A portion of the $H_2O_2$ diffuses to the surface of the first working electrode and is detected due to the transfer of two electrons to the electrode. Another portion of the $H_2O_2$ can diffuse out of the membrane. Since the diffusion barrier ***D*** is disposed between the working electrodes, the diffusion barrier substantially blocks diffusion of $H_2O_2$ to the second working electrode **E2.** If no diffusion barrier were present, the $H_2O_2$ would be able to diffuse to the second working electrode **E2** and cause a signal also referred to as cross talk. A variety of diffusion barriers can be employed to prevent cross talk. In some embodiments, the diffusion barrier ***D*** is a physical diffusion barrier, such as a structure between the working electrodes that blocks glucose and $H_2O_2$ from diffusing from the first working electrode **E1** to the second working electrode **E2.** In other embodiments, the diffusion barrier ***D*** is a spatial diffusion barrier, such as a.distance between the working electrodes that blocks glucose and $H_2O_2$ from diffusing from the first working electrode **E1** to the second working electrode **E2.** In still other embodiments, the diffusion barrier ***D*** is a temporal diffusion barrier, such as a period of time between the activity of the working electrodes such that if glucose or $H_2O_2$ diffuses from the first working electrode **E1** to the second working electrode **E2,** the second working electrode **E2** will not substantially be influenced by the $H_2O_2$ from the first working electrode **E1**.

**[0505]** Accordingly, in some preferred embodiments, the dual-electrode sensor comprises an insulator, such as an electrical insulator, located between the first and second working electrodes, wherein the insulator comprises a physical diffusion barrier. The physical diffusion barrier is configured to structurally block a substantial amount of diffusion of at least one of an analyte (e.g., glucose) and a co-analyte (e.g., $H_2O_2$) between the first and second working electrodes. In some embodiments, the diffusion barrier comprises a structure that protrudes from a plane that intersects both the first and second working electrodes. In a further embodiment, the structure that protrudes comprises an electrical insulator and/or an electrode.

**[0506]** In some preferred embodiments, the dual-electrode sensor comprises an insulator located between the first and second working electrodes, wherein the insulator comprises a diffusion barrier configured to substantially block diffusion of at least one of an analyte and a co-analyte between the first and second working electrodes. In preferred embodiments, the diffusion barrier comprises a temporal diffusion barrier configured to block or avoid a substantial amount of diffusion or reaction of at least one of the analyte and the co-analyte between the first and second working electrodes.

**[0507]** In still other preferred embodiments, the dual-electrode sensor comprises a sensor membrane configured to substantially block diffusion of at least one of an analyte and a co-analyte between the first and second working electrodes by a discontinuity of the sensor membrane between the first and second working electrodes. A discontinuity of the sensor

membrane is a type of physical diffusion barrier formed by a portion of the membrane between the two working electrodes, in some embodiments, wherein a discontinuity in the membrane structure blocks diffusion of $H_2O_2$ between the electrodes. Discontinuities of sensor membranes are discussed in greater detail with reference to **Fig. 3I**, in the section entitled "Sensor Configurations for Equivalent Measurement of Noise."

**[0508]** In some embodiments, the dual-electrode sensor system is configured for fluid communication with a host's circulatory system, such as *via* a vascular access device. A variety of vascular access devices suitable for use with a dual-electrode analyte sensor are described elsewhere herein. In some embodiments, the vascular access device comprises a lumen and at least a portion of the sensor is disposed within the lumen; and in some embodiments, at least a portion of the sensor can extend into the vascular system. In some embodiments, the vascular access device comprises a hub and the continuous analyte sensor is disposed substantially within the hub. In some embodiments, the system includes a fluid coupler configured and arranged to mate with the vascular access device on a first end; wherein the sensor is disposed within a portion of the fluid coupler and/or at a surface of the fluid coupler. In some embodiments, the sensor is configured to reside substantially above a plane defined by the host's skin. In some embodiments, the sensor is disposed on a surface of the vascular access device. In some embodiments, the vascular access device is configured for insertion into at least one of an artery, a vein, a fistula, and an extracorporeal circulatory device configured to circulate at least a portion of the host's blood outside of the host's body. In some embodiments, the system includes a flow control device in fluid communication with the vascular access device. The flow control device is configured to meter a flow of a fluid (e.g., blood, saline, a reference solution) through the vascular access device. In some embodiments, the flow control device is further configured to control fluid contact with the continuous analyte sensor, as is described in the section entitled "Integrated Sensor System."

**[0509]** In preferred embodiments, the sensor electronics (e.g., electronic components) are operably connected to the first and second working electrodes. The electronics are configured to calculate at least one analyte sensor data point. For example, the electronics can include a potentiostat, A/D converter, RAM, ROM, transmitter, and the like. In some embodiments, the potentiostat converts the raw data (e.g., raw counts) collected from the sensor to a value familiar to the host and/or medical personnel. For example, the raw counts from a glucose sensor can be converted to milligrams of glucose per deciliter of glucose (e.g., mg/dl). In some embodiments, the electronics are operably connected to the first and second working electrodes and are configured to process the first and second signals to generate a glucose concentration substantially without signal contribution due to non-glucose noise artifacts. The sensor electronics determine the signals from glucose and non-glucose related signal with an overlapping measuring potential (e.g., from a first working electrode) and then non-glucose related signal with an overlapping measuring potential (e.g., from a second electrode). The sensor electronics then use these data to determine a substantially glucose-only concentration, such as but not limited to subtracting the second electrode's signal from the first electrode's signal, to give a signal (e.g., data) representative of substantially glucose-only concentration, for example. In general, the sensor electronics may perform additional operations, such as but not limited to data smoothing and noise analysis.

**[0510]** In preferred embodiments, the dual-electrode sensor includes electronics (e.g., a processor module, processing memory) that are operably connected to the first and second working electrodes and are configured to provide the first and second signals to generate an analyte concentration data substantially without signal contribution due to non-analyte-related noise. For example, the sensor electronics process and/or analyze the signals from the first and second working electrodes and calculate the portion of the first electrode signal that is due to analyte concentration only. The portion of the first electrode signal that is not due to the analyte concentration can be considered to be background, such as but not limited to noise. Accordingly, in one embodiment of a dual-electrode sensor system configured for fluid communication with a host's circulatory system (e.g., *via* a vascular access device) the system comprising electronics operably connected to the first and second working electrodes; the electronics are configured to process the first and second signals to generate analyte concentration data substantially without signal contribution due to noise.

**[0511]** In some embodiments, the dual-electrode analyte sensor includes a reference sensor/system, as described elsewhere therein, whereby reference data can be provided for calibration (e.g., internal to the system), without the use of an external (e.g., separate from the system) analyte-measuring device. In an exemplary embodiment, external glucose data points (e.g., from a hand-held glucose meter or a YSI device) are not required for calibration of a dual-electrode glucose sensor system that includes a reference sensor. In some embodiments, the reference sensor is configured to be disposed within the same local environment as the dual-electrode analyte sensor, such that the reference sensor and the dual-electrode analyte sensor can be simultaneously exposed to a sample. In some embodiments, the reference sensor/system can be disposed remotely from the dual-electrode sensor. In these embodiments, the electronics module is configured to process the reference data with the first and second signals to generate analyte concentration data substantially without signal contribution due to noise. In some embodiments, the electronics module is configured to calibrate the dual-electrode analyte sensor data using the reference sensor data, as described elsewhere herein.

**[0512]** In some embodiments, the electronics module is configured to determine a scaling factor ($k$) as described in the section entitled "Calibration Systems and Methods." Briefly, a scaling factor defines a relationship between the enzymatic portion of the membrane and the non-enzymatic portion of the membrane. Accordingly, in some embodiments,

the electronics module, also referred to as the processor module herein, is configured to calibrate the analyte sensor data using the scaling factor, such that the calibrated sensor data does not include inaccuracies that can arise due to small differences between the plus- and minus-enzyme portions of the membrane at the first and second working electrodes, respectively.

**[0513]** In some embodiments, the system is configured to calibrate the continuous dual-electrode analyte sensor using a reference fluid (e.g., **602a**), as described in the section entitled "integrated sensor system." In some embodiments, the system is configured to calibrate the sensor using single-point calibration, in other embodiments, the system is configured to calibrate the sensor without a reference data point provided by an external analyte monitor (e.g., SMBG, YSI), as described elsewhere herein. In some embodiments, the system includes a reference sensor configured to generate a signal associated with a reference analyte in the sample (e.g., internal to the system), wherein the continuous analyte sensor is further configured to generate a third signal associated with the reference analyte, and wherein the system is configured to calibrate the continuous analyte sensor using the reference signal and the third signal. In some embodiments, the reference sensor comprises an optical sensing apparatus, such as but not limited to an optical $O_2$ sensor. In preferred embodiments, the continuous analyte sensor is a glucose sensor. In other embodiments, a substantial portion of the continuous analyte sensor has a diameter of less than about 0.008 inches, as is described elsewhere herein.

**[0514]** In some further embodiments, the continuous analyte sensor further comprises a bioinert material or a bioactive agent incorporated therein or thereon. Applicable bioactive agent include but are not limited to vitamin K antagonists, heparin group anticoagulants, platelet aggregation inhibitors, enzymes, direct thrombin inhibitors, Dabigatran, Defibrotide, Dermatan sulfate, Fondaparinux, and Rivaroxaban.

**[0515]** As a non-limiting example, in some preferred embodiments, a method for continuously detecting an analyte in the host *in vivo* using a dual-electrode analyte sensor is provided. In some embodiments, a vascular access device (e.g., a catheter) is inserted into the host's circulatory system, such as into a vein or artery. The sensor is contacted with a sample of the circulatory system, such as a sample of blood withdrawn into the catheter. A first signal is generated by the sensor, wherein the first signal is associated with associated with the analyte and non-analyte related electroactive compounds having a first oxidation/reduction potential in a sample of the circulatory system of the host. In preferred embodiments, the analyte sensor is configured to detect glucose. A second signal is also generated, wherein the second signal is associated with noise of the analyte sensor, wherein the noise comprises signal contribution due to non-analyte related electroactive species with an oxidation/reduction potential that substantially overlaps with the first oxidation/reduction potential in the sample. The first and second signals are processed to provide a processed signal substantially without a signal component associated with noise. In some embodiments, the first and second signals are processed to provide a scaling factor, which can then be used to calibrate the first signal. In some embodiments, a reference sensor is also contacted with the sample, and a third signal associated with a reference analyte generated. In some embodiments, the reference sensor is an optical detection apparatus, such as but not limited to an optical $O_2$ sensor. In this embodiment, the first and second signals can be calibrated using the third and/or reference signal. In some preferred embodiments, the processing step comprises evaluating steady-state information and transient information, wherein the first and second signals each comprise steady-state and transient information. In some further embodiments, the evaluating step includes evaluating at least one of sensitivity information and baseline information, wherein the steady-state information comprises the sensitivity and baseline information.

Optical Detection

**[0516]** In some embodiments, the continuous analyte sensor is configured to detect the analyte by optical means. In some embodiments, various types of Raman and/or fluorescent spectroscopic detection are used. For example, glucose can be detected *via* fiber optic visible fluorescence, using glucose dehydrogenase (GDH) and a modified flavin adenine dinucleotide (FAD) coenzyme system, Concanavalin A, or hexokinase. In some embodiments, a fluorescent molecule (e.g., a fluorophore) is attached to the co-enzyme or to a hydrogen peroxide end-product reactant, as in a colorimetric detection system. In an alternative embodiment, ferrocene is modified and used as a mediator in the reaction of GOX (or GHD) with glucose, wherein a pH-sensitive fluorophore is used to detect the reaction. In some embodiments, fiber optic probes are constructed by applying membrane systems configured for optical/fluorescent detection to an optical fiber. In some embodiments, multiple fiber optic probes are bundled, which enables a variety of integrative and/or subtractive signal correlations/corrections and to enhance error detection. Examples of optical detection can be found in U.S. Patent Nos. 7,289,836 and 7,149,562.

**[0517]** In some embodiments, a continuous analyte detection system is provided, including a sensor configured and arranged for fluid contact with a host's circulatory system and a processor module. The sensor comprises both a continuous analyte sensor (e.g., either non-dual-electrode or dual-electrode) and a reference sensor. For example, in some embodiments the system includes a continuous analyte sensor including a working electrode and a reference electrode, and a reference sensor. In other embodiments, the system includes a dual-electrode analyte sensor, including first and second working electrodes and a reference electrode, and a reference sensor. The continuous analyte sensor is con-

figured and arranged to generate a first signal associated with a test analyte and a second signal associated with a reference analyte. For example, in one embodiment, the test analyte is glucose and the reference analyte is oxygen; thus, the first signal is associated with glucose and the reference signal is associated with oxygen. The reference sensor is configured to generate a reference signal that is also associated with the reference analyte. In general, a "reference analyte" can be any analyte that can be measured by both the analyte sensor and the reference sensor, such those analytes listed under the definition of "analyte" in the section entitled "Definitions." In preferred embodiments, the reference analyte is one that is relatively stable within the host's body, such as but not limited to $O_2$, succinate, glutamine, and the like. In this embodiment, the processor module is configured to process the second signal (e.g., related to the reference analyte) and the reference signal to calibrate the first signal (e.g., related to the analyte). In some embodiments, the processor module calibrated the second signal (e.g., the reference analyte signal detected by the analyte sensor) using the reference signal provided by the reference sensor, and then to calibrate the first signal (e.g., the analyte signal) using the second signal.

**[0518]** As a non-limiting example, in some embodiments, the system's continuous analyte sensor is a dual-electrode sensor that comprises both first and second working electrodes **E1, E2.** Accordingly, the first working electrode is disposed beneath an active enzymatic portion of a sensor membrane and generates a signal (e.g., the first signal) associated with both the analyte (e.g., glucose) and non-analyte related electroactive compounds (e.g., non-glucose compounds that have an oxidation/reduction potential that substantially overlaps with the oxidation/reduction potential of glucose). Additionally, the second working is disposed beneath an inactive-enzymatic or a non-enzymatic portion of the sensor membrane and generates a non-analyte-related signal associated with the non-analyte electroactive species. In this embodiment, the processor module is configured to process signals from the first and second working electrodes, and to thereby generate a first signal substantially without a non-analyte signal component.

**[0519]** The second signal (e.g., related to the reference analyte) can be generated by various means. For example, in some embodiments, the first working electrode of the dual-electrode analyte sensor is configured to generate both the first signal and the second signal. For example, in some embodiments, pulsed amperometric detection, including switching, cycling or pulsing the voltage of the electrode in an electrochemical system (e.g., between a positive voltage (e.g., +0.6 for detecting glucose) and a negative voltage (e.g., -0.6 for detecting oxygen)) can be employed to determine an oxygen measurement. For example, the first working electrode is configured to generate a signal associated with the analyte when a potential of +0.6mV is applied thereto. If the potential is switch to -0.6mV, then the first working electrode becomes an $O_2$ sensor and measures a signal associate with the amount of $O_2$ passing through the sensor's membrane system. U.S. Patent No. 4,680,268 to Clark, Jr. described pulsed amperometric detection in greater detail. Additional oxygen sensors are described in U.S. Patent No. 6,512,939 to Colvin.

**[0520]** As a non-limiting example, in one embodiment the dual-electrode analyte sensor is a glucose sensor configured for fluid communication with a host's circulatory system, wherein the sensor is configured to generate a first signal associated with glucose (at the first working electrode **E1**) at an applied potential of 0.6mV, and then to generate a second signal associated with $O_2$ (also at the first working electrode **E1**) at an applied potential of -0.6mV. Thus, in some embodiments, the potential applied to the first working electrode can be switched from +0.6mV to -0.6mV, such that the first working electrode switches from measuring the analyte-related signal (e.g., glucose) to measuring the second signal (e.g., associated with $O_2$).

**[0521]** In some alternative embodiments, the second working electrode **E2** (e.g., instead of the first working electrode **E1**) is configured to generate the second signal. As a non-limiting example, in another embodiment the dual-electrode analyte sensor is a glucose sensor configured for fluid communication with a host's circulatory system, wherein the sensor is configured to generate a first signal associated with glucose (at the first working electrode **E1**) at an applied potential of 0.6mV, to generate a non-analyte-related signal (at the second working electrode **E2**) at an applied potential of 0.6mV, and then to generate a second signal associated with $O_2$ (also at the second working electrode **E2**) at an applied potential of -0.6mV. Thus, in some embodiments, the potential applied to the second working electrode can be switched from 0.6mV to -0.6mV, such that the second working electrode switches from measuring the non-analyte-related signal to measuring the second signal (e.g., associated with $O_2$).

**[0522]** In still other embodiments, the second signal can be generated by a third working electrode disposed beneath the sensor's membrane. As a non-limiting example, in another embodiment the dual-electrode analyte sensor is a glucose sensor configured for fluid communication with a host's circulatory system, wherein the sensor is configured to generate a first signal associated with glucose (at the first working electrode **E1**) at an applied potential of 0.6mV, to generate a non-analyte-related signal (at the second working electrode **E2**) at an applied potential of 0.6mV, and then to generate the second signal associated with $O_2$ (at the third working electrode, e.g., **E3,** not shown) at an applied potential of -0.6mV. Thus, in some embodiments, switching the applied potential from 0.6mV to -0.6mV is not required.

**[0523]** As described above, the system includes a reference sensor. In some embodiments, the reference sensor is an optical sensing apparatus, as described above. In other embodiments, the reference sensor is configured to detect the reference analyte by any means known in the art, such as but not limited to electrochemical, chemical, physical, immunochemical, calorimetric and/or radiometric means. Preferably, the reference sensor is disposed in the same local

environment as the continuous analyte sensor, but not under the membrane system of the continuous analyte sensor. For example, the reference sensor can be disposed adjacent to the continuous analyte sensor, such that when the continuous analyte sensor is contacted with a sample the reference sensor is simultaneously contacted by the sample. As a non-limiting example, in some embodiments, a dual-electrode continuous analyte sensor and a reference sensor are disposed adjacently, such that they can be simultaneously exposed to a sample of the host's circulatory system.

**[0524]** As a non-limiting example, in some embodiments, a dual-electrode sensor includes a first working electrode configured to detect the analyte (including non-analyte-related noise) and a second working electrode is configured to detect the signal associated with non-analyte-related noise. In some embodiments, the first and second working electrodes are bundled and/or twisted, and the reference sensor is disposed adjacent to the first and second working electrodes. In some embodiments, either the first or the second working electrodes of the dual-electrode sensor is configured to detect a signal associated with the reference analyte (e.g., a second signal); while in other embodiments, the dual-electrode sensor includes a third working electrode configured to detect the reference analyte (e.g., the second signal).

**[0525]** As a non-limiting example, in one embodiment of a system comprising both a dual-electrode sensor and an optical reference sensor, the dual-electrode sensor is both a glucose sensor and an $O_2$ sensor, and the reference sensor is an optical $O_2$ sensor. Accordingly, as described elsewhere herein, the first working electrode of the dual-electrode sensor is configured to detect a signal associated with both glucose and non-glucose-related electroactive species (in a sample of the host's circulatory system), and the second working electrode is configured to detect the non-glucose related electroactive species. Either the first working electrode or the second working electrode is configured to detect $O_2$, such as *via* switching the applied potential from 0.6mV to -0.6MV, as described elsewhere herein. The reference sensor, which can be bundled with the dual-electrode sensor, is configured to optically detect a reference signal associated with the $O_2$ concentration of the sample. In some embodiments, instead of using the first or second working electrodes to detect $O_2$, the dual-electrode sensor includes a third electrode configured to detect $O_2$.

**[0526]** In preferred embodiments, the signal related to a reference analyte (e.g., $O_2$) can be used to calibrate the signals from a continuous analyte sensor, such as in the event of a drift in sensor sensitivity and/or baseline. Accordingly, the signals related to the reference analyte, from the continuous analyte and reference sensors, can be processed to determine a calibration factor. The calibration factor can then be used to calibrate the continuous analyte sensor data. As used herein, the term "calibration factor" as used herein is a broad term, and is to be given its ordinary and customary meaning to a person of ordinary skill in the art (and is not to be limited to a special or customized meaning), and refers without limitation to a mathematical function generated by processing the reference analyte-related signal of the continuous analyte sensor and the reference analyte-related signal of the reference sensor, which can be used to calibrate the continuous analyte sensor data initially and/or responsive to an occurrence of a drift in sensor sensitivity and/or baseline.

**[0527]** In some embodiments, a method for measuring an analyte in a host is provided. First, a continuous analyte detection system, which includes a continuous analyte sensor and a reference sensor, is provided. In preferred embodiments, the continuous analyte sensor is configured and arranged to generate a first signal associated with a test analyte (e.g., glucose) and a second signal associated with a reference analyte (e.g., $O_2$), and the reference sensor configured to generate a reference signal associated with the reference analyte (e.g., $O_2$).

**[0528]** Next, the detection system is exposed to a sample of a host's circulatory system *in vivo.* For example, the detection system can be fluidly coupled to a vascular access device implanted in a host's circulatory system, such that a sample of the host's blood can be drawn back an contacted with the detection system. Preferably, the continuous analyte sensor and the reference sensor are exposed to the sample simultaneously. Accordingly, measurements of the reference analyte can be made at the same time by the continuous analyte and reference sensors. In some embodiments, a fluid flow device configured for fluid communication with the circulatory system of the host and to meter a flow of a fluid therethrough is provided. In these embodiments, the fluid flow device comprises a vascular access device configured for insertion into either an artery or a vein of the host. Such fluid flow devices are described in detail in the sensor entitled "Integrated Sensor System." In preferred embodiments, the fluid flow device is coupled with the continuous analyte detection system, and a sample of the circulatory system of the host is withdrawn. In some preferred embodiments, the fluid flow device is further configured to meter the flow of a non-bodily fluid through the vascular access device. Non-bodily fluids include a variety of sterile infusion fluids, such as but are not limited to saline, reference solutions such as a glucose solution of defined concentration, nutritional supplements, IV medicaments, and the like.

**[0529]** When the sample contacts the system, signals are received from the continuous analyte sensor and the reference sensor. The signals received include a first signal (e.g., related to the test analyte, such as but not limited to glucose), a second signal (e.g., related to the reference analyte, such as but not limited to $O_2$) and a reference signal (e.g., related to the reference analyte). In some embodiments, the first signal is received from a first working electrode disposed under an enzymatic portion of a membrane system. For example, in the case of a glucose detection system, the first working electrode is disposed beneath a portion of the membrane system including active GOX and detects a first signal associated with the concentration of glucose in the sample. In some embodiments, the first working electrode

also received the second signal, as described elsewhere herein. In other embodiments, the second signal is received from a second working electrode that is also disposed under the membrane system. In some embodiments, the second working electrode configured to also receive a non-analyte-related signal. For example, the second working electrode is disposed under a non-enzymatic portion of the membrane system, in some embodiments. In some other embodiments, the non-analyte-related signal is received from a third working electrode disposed under a non-enzymatic portion of the membrane system. In some embodiments, the reference sensor is configured to detect the reference analyte optically. For example, in some embodiments, the reference analyte is oxygen. Accordingly, the second signal and the reference signal received are associated with the concentration of oxygen in the sample.

**[0530]** After the signals have been received, a calibration factor is calculated, wherein the calibration factor is associated with a sensitivity and/or baseline of the continuous analyte sensor. For example, in some embodiments, the continuous analyte detection system is exposed to a bodily fluid (e.g., blood) and the calculating step includes comparing steady-state information of the first signal and steady-state information of the second signal. In some embodiments, the calibration factor can be calculated by examining the transient information of the first and second signals.

**[0531]** In some other embodiments, the continuous analyte detection system is configured to be exposed to a non-bodily fluid, such as saline or a reference fluid, such as to wash the previous blood sample off of the device. During the washing procedure, the non-bodily fluid can be held substantially stagnant (e.g., no flow or very little flow of the fluid past the sensor) for a period of time. During this period of time, the working electrodes of the continuous analyte sensor detect signals associated with non-analyte-related compounds diffusion to the first and second working electrodes. For example, in some embodiments, a saline solution containing a defined amount of glucose is used to wash the sensor. When the glucose-containing saline is held stagnant (e.g., after washing the previous sample off of the sensor), the GOX at the first working electrode metabolizes glucose diffusing through the membrane. As the glucose is metabolized, the reactant (e.g., $H_2O_2$) begins to accumulate and produce signals at both the first and second working electrodes. As a result, the signal increase on each of the first and second working electrodes (e.g., during the time period) can be compared to calculate the calibration factor. This method for calculating the calibration factor is discussed in greater detail elsewhere herein, with reference to **Figs. 3J** and **3K.**

**[0532]** After the calibration factor has been calculated, the signal(s) from the continuous analyte sensor are calibrated using the calibration factor. The process of calculating the calibration factor and then using the newly calculated calibration factor to calibrate the signals can be continuous, continual and/or intermittent; such that at time passes the calibration factor and calibration are updated. Thus, the system is configured to evaluate changes in membrane sensitivity and/or baseline, to adjust the calculation of analyte concentrations accordingly, whereby the host is provided with more accurate data for use in therapy decision-making.

Multi-sensor Apparatus

**[0533]** In some preferred embodiments, a multi-sensor apparatus configured for the detection of a plurality of analytes in a circulatory system of a host *in vivo* is provided. **Figs. 2G** through **2L** illustrate some exemplary embodiments of such a device. In prefered embodiments, the multi-sensor apparatus is a vascular access device (e.g., a catheter) or a connector configured for fluid communication with the circulatory system of the host. Preferably, the multi-sensor apparatus includes a lumen (e.g., a duct) sufficiently large to house the plurality of sensors, as described elsewhere herein. In an exemplary embodiment, the multi-sensor apparatus comprises a plurality of analyte sensors, wherein the plurality of analyte sensor are configured to detect at least one analyte and to contact a sample of the host's circulatory system. In one exemplary embodiment, the multi-sensor apparatus comprises a lumen, an external surface, and two orifices, wherein a first orifice is proximal relative to the host and the second orifice is distal. In some embodiments, such as in a catheter, the proximal orifice is referred to herein as the *in vivo* orifice and the distal orifice is referred to as the *ex vivo* orifice. Preferably, at least the distal orifice is configured to couple with a fluid flow device (or a component thereof), such as but not limited to a connector or coupler, a valve, IV tubing, a pump, and the like. For example, in an embodiment wherein the multi-sensor apparatus is a catheter, the distal orifice (e.g., the ex vivo orifice) is configured to couple to IV tubing, various types of IV connectors, and the like. In some embodiments, both the proximal and distal orifices are configured to couple with IV equipment. For example, in an embodiment wherein the multi-sensor apparatus is configured as a connector (e.g., a Leur lock) the proximal orifice is configured to couple with a vascular access device (e.g., a catheter/cannula), IV tubing, and/or other connectors, and the distal end is configured to couple with a fluid flow device (e.g., IV tubing, a pump, etc.). Preferably; a plurality of analyte sensors are disposed within the lumen of the multi-sensor apparatus. For example, 2, 3, 4, 5, 6, 7, or more sensors can be disposed within the lumen of the multi-sensor apparatus. In some embodiments, each of the plurality of analyte sensor is configured to detect a different analyte. In some embodiments, two or more of the plurality of analyte sensors are configured to detect the same analyte, thereby providing redundancy and/or fail-safes in analyte detection and/or sensor function.

**[0534]** **Fig. 2G** provides an exemplary embodiment of a multi-sensor apparatus, namely a catheter, including an *in vivo* portion configured for insertion into the host and an *ex vivo* portion **218** (e.g., a connector or hub) configured to

remain outside the host's body after implantation/insertion of the *in vivo* portion into a host. The *in vivo* portion may also be referred to as the proximal portion/end of the catheter (e.g., with respect to the host) includes an *in vivo* orifice at or near the catheter's tip, for fluid communication with the host's circulatory system upon implantation into the host's vein or artery, or in an extracorporeal circulatory device. The *ex vivo* portion of the catheter may also be referred to as the proximal portion (e.g., with respect to the host). A plurality of analyte sensors **240** are disposed within the catheter's connector/hub, such as within the lumen/duct **254** and/or within a widened portion of the catheter's *in vivo* portion.

**[0535]** **Fig. 2H** provides another exemplary embodiment of a multi-sensor apparatus, namely a connector, such as a Leur lock, a Y-connector, a T-connector, an X-connector, or a valve configured for connecting IV equipment. The multi-sensor apparatus includes a proximal orifice (e.g., with respect to the host) configured to couple with a vascular access device (e.g., a catheter/cannula) or with various IV equipment, such as IV tubing or another connector, and a distal orifice (e.g., with respect to the host) configured to fluidly couple to other IV equipment, as described herein and is known to one skilled in the art. The analyte sensors **240** are disposed within the multi-sensor apparatus's lumen/duct **254.**

**[0536]** In various embodiments, the analyte sensors (of the multi-sensor apparatus) can be configured to detect an analyte using any means known in the art, such as but not limited to by enzymatic, chemical, physical, electrochemical, spectrophotometric, polarimetric, calorimetric, radiometric, or immunochemical techniques, or by a combination of these techniques. Further more, each sensor can use a different detection technique. For example, a first analyte sensor can detect a first analyte using a first technique, a second analyte sensor can detect a second analyte using a second technique, a third analyte sensor can detect a third analyte using a third technique, and so on. Additionally, in some embodiments, a detection technique can be used by more than one of the analyte sensors, wherein the technique is modified to detect a particular analyte of interest by each of the sensors. For example, a first sensor can be configured to detect glucose enzymatically, and a second sensor can be configured to detect cholesterol enzymatically. In some embodiments, one of the plurality of sensors is configured to detect an analyte optically, as described elsewhere herein. Additionally, in some embodiments, two or more of the sensors are configured to detect the same analyte, either by the same or different detection techniques.

**[0537]** **Figs. 2I** through **2L** are cross-sections of the multi-sensor apparatus of **Figs. 2G** and **2H** taken along line **2I - 2I**, looking towards the proximal ends (e.g., **212b/258**) of the devices. A plurality of analyte sensors **240** is disposed at the luminal surface of wall **260** (e.g., the interior surface of the hub/connector). In some embodiments, one or more of the plurality of sensors is integrally formed with the multi-sensor apparatus. In some embodiments, the multi-sensor apparatus includes a plurality of sensor sites **262,** wherein each sensor site **262** is configured for the disposition of a sensor. In some embodiments, at least one of the plurality of sensor sites **262** comprises a breakaway portion (or a plug) configured for insertion therethrough of a sensor, such that at least a portion of the sensor is disposed within the lumen. One or more of the breakaway portions can be removed, such a by punching them out, to form a channel through the wall **260.** In some embodiments, the multi-sensor apparatus is manufactured such that one or more of the sensor sites includes a channel (e.g., through the wall), such that a sensor can be inserted there through. The sensor(s) can be installed by insertion through the channel(s). An adhesive, press-fit, clip or other attachment means can be use to secure the sensor(s) in place. In some embodiments, a portion of a sensor **240** (e.g., the sensing portion) inserted through the wall **260** is disposed at the surface of the duct/lumen. In some embodiments, the portion of the sensor protrudes into the duct/lumen **254.** In some further embodiments, at least another portion of the sensor is disposed at the external surface of the connector/hub. In some embodiments, one or more sensors can be disposed (e.g., installed) within the duct/lumen by adhering the sensor at the surface of the duct/lumen. In some embodiments, one or more of the sensors is deposited at the surface of the duct/lumen using known analyte sensor deposition techniques. In some embodiments, conductive traces, leads or wires can be applied/installed, such that the sensor(s) can be connected to device electronics, as is understood by one skilled in the art. For example, the device shown in **Figs. 1A** and **1B** include a conductive lead **24,** for connecting the analyte sensor to electronics.

**[0538]** Referring again to **Fig. 2G,** in some embodiments, the multi-sensor apparatus is a vascular access device comprising an *in vivo* portion and an *ex vivo* portion. In some preferred embodiments, the plurality of analyte sensors are disposed only within the *ex vivo* portion of the device, and thus do not extend into the *in vivo* portion (e.g., catheter **212**). In this embodiment, the plurality of sensors does not extend beyond a plain defined by the host's skin. In some embodiments, the *in vivo* portion of the multi-sensor apparatus includes a widened portion, such as a portion adjacent to and/or near to the hub, and one or more of the plurality of sensors are disposed within the widened portion. In some embodiments, one or more of the analyte sensor can be configured to extend into the *in vivo* portion, and in some embodiments to extend into the host's circulatory system.

**[0539]** Referring again to **Fig. 2H,** in some embodiments, the multi-sensor apparatus is a connector configured to be disposed outside the host's body. Accordingly, the multi-sensor apparatus does not include an *in vivo* portion. In this embodiment, the multi-sensor apparatus is configured to fluidly couple to a vascular access device at its proximal end and to a flow control device at its distal end, such that the flow control device can meter the flow of a non-bodily fluid (e.g., saline, a glucose solution, etc.) through the device and into the host, as well as withdrawal of blood samples from the host (e.g., such that the sample(s) contact the analyte sensor(s)) and (optionally) reinfusion of the blood samples to

the host. The multi-sensor apparatus of this embodiment includes a lumen and/or duct, in which the plurality of analyte sensors is disposed. In some embodiments, at least one of the plurality of analyte sensors is configured to extend into the lumen of a fluidly coupled catheter; and in some further embodiments to extend through the catheter and into the host's s circulatory system.

**[0540]** In preferred embodiments, at least one of the plurality of sensors (of the multi-sensor apparatus) is configured to generate a signal associated with a concentration of an analyte in a sample of the host's circulatory system. More preferably, each of the analyte sensors generates a signal associated with a concentration of each sensor's respective analyte in the blood sample withdrawn from the host. In some embodiments, the sensors can be configured to generate signals associated with 1, 2, 3, 4, 5, 6, 7, 8, 9, 10 or more analytes and/or properties of the sample of the host's circulatory system. In some embodiments, two or more of the sensors are configured to detect the same analyte, such as to provide system redundancy and/or fail-safes. The analyte sensors can be configured to detect a wide variety of analytes, such as but not limited to glucose, oxygen, $CO_2$ (carbon dioxide, bicarbonate), pH, creatinine, urea (nitrogen), bilirubin, electrolytes (e.g., sodium, potassium, chloride, phosphorous, magnesium), albumin, total protein, liver enzymes (e.g., alkaline phosphatase, alanine amino transferase, aspartate amino transferase), antibodies against infective agents, fibrinogen, fibronectin, lipids, triglycerides, cholesterol-protein complexes and ratios thereof (e.g., LDL, HDL, chylomycrons), hormones (e.g., T3, T4, TSH, hGH, interleukins, etc.), medicaments, metabolites, and the like. A more extensive list of analytes can be found in the "Definitions" section. In some embodiments, at least one of the plurality of sensors is configured to generate a signal associated with a property of a sample of the host's circulatory system. Blood properties include but are not limited to pH, temperature, oxygen tension, hematocrit, viscosity, clotting, pressure, and the like.

**[0541]** The multi-sensor apparatus of the preferred embodiments can be manufactured using a variety of techniques known in the art. For example, in some embodiments, the analyte sensors are integrally formed with the multi-sensor apparatus. In some embodiments, at least one of the plurality of sensors is deposited within the lumen of the multi-sensor apparatus, such as in the lumen/duct of the connector of the hub of the device illustrated in **Fig. 2G,** or in the lumen/duct of the device of **Fig. 2H.** In some embodiments, one or more of the analyte sensors is configured to extend out of the connector/hub. For example, in the exemplary embodiment illustrated in **Fig. 2G** one or more analyte sensors **240** can be configured to extend into and/or through the lumen **212a** of the catheter **212.** In another example, in the exemplary embodiment illustrated in **Fig. 2H** one or more analyte sensors **240** can be configured to extend out of the proximal end of the multi-sensor apparatus, such that the sensor(s) can be inserted into and/or through a vascular access device.

**[0542]** In some embodiments, the non-sensor portion of a multi-sensor apparatus is formed, and then the plurality of sensors are applied/installed. In some embodiments, at least one of the plurality of sensors is deposited within the lumen, such as by screen printing. In some embodiments, at least one of the plurality of sensors is applied to the interior surface of the lumen, such as *via* an adhesive.

**[0543]** Alternatively, the multi-sensor apparatus may be formed about the plurality of analyte sensors, such as by using injection molding. For example, a mold is prepared, including sites for the analyte sensors (e.g., these will be "sensor sites," as described elsewhere herein, when the manufacture process is complete). Prior to injection molding, the sensors (e.g., previously manufactured) are placed in the sites. The mold is closed and a material, such as but not limited to, e.g., molten plastic, is injected into the mold. The material fills all of the spaces within the mold, including flowing around portions of the analyte sensors, such that when the mold is cooled, the analyte sensors will be held in place by the wall of the multi-sensor apparatus. For example, one or more of the sensors pass through the wall of the multi-sensor apparatus. In another example, the sensor can be oriented such that when the injection molding process is completed, the analyte sensor is disposed on the surface of the lumen. In some embodiments, one or more of the analyte sensors can be installed in the multi-sensor apparatus during injection molding, followed by application of one or more additional sensors to the lumen of the device.

**[0544]** As a non-limiting example, a method for making a multi-sensor apparatus for the detection of a plurality of analytes in a circulatory system of a host *in vivo* is provided, in some embodiments. In some embodiments, a plurality of sensors are first provided. The sensors can be configured to detect one or more analytes using a variety of detection means known to one skilled in the art. Next, a multi-sensor apparatus is formed about the plurality of sensors. The multi-sensor apparatus formed includes a lumen, an external surface, and at least one orifice configured for coupling with a fluid flow device, as described herein.

**[0545]** As another non-limiting example, in some embodiments, a method for making a multi-sensor apparatus for the detection of a plurality of analytes in a circulatory of a host *in vivo* includes providing a multi-sensor apparatus comprising a lumen, an external surface, and at least one orifice configured for coupling with a fluid flow device a plurality of sensors; followed by forming a plurality of sensor within and/or on the multi-sensor apparatus.

**[0546]** As yet another non-limiting example, a method for detecting of a plurality of analytes in a circulatory of a host *in vivo,* using a multi-sensor apparatus of the preferred embodiments, is provided. Accordingly, a multi-sensor apparatus of the preferred embodiments is applied to the circulatory system of a host. As described elsewhere herein, the multi-sensor apparatus includes a lumen and a plurality of sensors, wherein the at least two sensor are disposed above a

plane defined by the skin of the host when the multi-sensor apparatus is applied to the host's circulatory system. For example, in some embodiments, the multi-sensor apparatus is a catheter with sensors in the hub/connector, which is inserted/implanted into a host's artery/vein. After the catheter has been inserted/implanted into the host, at least two of the sensors remain disposed outside the host's body, as defined by the host's skin. As another example, in some embodiments, the multi-sensor apparatus is a connector with sensors within its lumen. In this embodiment, the multi-sensor apparatus must be fluidly coupled to a vascular access device, so that blood can be withdrawn from the host's artery/vein and then contact the sensors within the connector. Thus, at least two of the sensors within this embodiment of the multi-sensor apparatus remain disposed outside the host's body, as defined by the host's skin.

**[0547]** Next, a sample (e.g., blood) is withdrawn from the host's circulatory system. When the sample is withdrawn, it is then contacted with the plurality of sensors. Each of the sensors then generates a signal. As described elsewhere herein, each sensor is configured to detect an analyte. Accordingly, the signal generated by each sensor is associated with the analyte that sensor was configured to detect. The sensors can be configured to generate the signal using any method know in the art, such as but not limited to electrochemically generating the signal, optically generating the signal, radiochemically generating the signal, physically generating the signal, chemically generating the signal, immunochemically generating the signal, and/or enzymatically generating a signal, or combinations thereof.

**[0548]** In some embodiments, a withdrawn sample is reinfused into the host. For example, a flow control device can meter the flow of an infusion fluid into the host, and the infusion fluid pushes the withdrawn sample back into the host. In some embodiments, the device is configured to dispose of the withdrawn sample, such as by directing the sample to a waste container.

**[0549]** As described herein, in some embodiments, an infusion fluid is metered through the multi-sensor apparatus, and infused into the host. In some embodiments, the plurality of sensors are washed with the infusion fluid. For example, infusion of about 0.5, 1, 5, 10, 15 ml or more of infusion fluid into the host can effectively wash a previous blood sample off of the plurality of analyte sensor in some embodiments. A variety of infusion fluids can be used, including but not limited to saline, reference fluids, medicaments, parenteral nutrition fluids, hydration fluid and the like.

**[0550]** In preferred embodiments, the signal of at least one of the plurality of sensors can be calibrated. A variety of calibration methods can be used. In some embodiments, one or more of the analyte sensors can be calibrated using one or more reference data points provided by a device separate from the multi-sensor apparatus/system. For example, a hand-held glucose meter can be used to provide one or more data points for calibrating a glucose sensor disposed in the multi-sensor apparatus. In some embodiments, a substantially stable and/or constant analyte found in the host's blood can be used to calibrate one or more of the plurality of analyte sensors. In some embodiments, data from a recently disconnected multi-sensor apparatus can be used to calibrate one or more of the sensors of a newly installed/applied multi-sensor apparatus. In some embodiments, one of the plurality of analyte sensors can be used to calibrate one or more of the other sensors. In some embodiments, one or more of the sensors can be calibrated by the manufacturer. Additional methods of calibration that can be used with the multi-sensor apparatus of the preferred embodiments are described elsewhere herein.

Noise

**[0551]** Generally, implantable sensors measure a signal (e.g., counts) related to an analyte of interest in a host. For example, an electrochemical sensor can measure glucose, creatinine, or urea in a host, such as an animal, especially a human. Generally, the signal is converted mathematically to a numeric value indicative of analyte status, such as analyte concentration. However, it is not unusual for a sensor to experience a certain level of noise. The term "noise" generally refers to a signal detected by the sensor that is substantially non-analyte related (e.g., non-glucose related). In other words, things other than the analyte concentration substantially cause noise. Noise is clinically important because it can reduce sensor performance, such as by making the analyte concentration appear higher or lower than the actual concentration. For example, if a host is hyperglycemic (e.g., blood sugar too high, greater than -120 mg/dl) or euglycemic (e.g., -80-120 mg/dl), noise can cause the host's blood sugar to appear higher than it truly is, which can lead to improper treatment decisions, such as to give the host an excessive insulin dose. An excessive insulin dose, in some circumstances, can lead to a dangerous hypoglycemic state (e.g., blood sugar too low, less than -80 mg/dl). In the case of a hypoglycemic host, noise can cause the hosts blood sugar to appear euglycemic or even hyperglycemic, which can also lead to improper treatment decisions, such as not eating when necessary or taking insulin, for example. Accordingly, since noise can cause error and reduce sensor performance, noise reduction is desirable.

**[0552]** Noise is comprised of two components, constant noise and non-constant noise, and can be caused by a variety of factors, ranging from mechanical factors to biological factors. For example, it is known that macro- or micro-motion, ischemia, pH changes, temperature changes, pressure, stress, or even unknown mechanical, electrical, and/or biochemical sources can cause noise. In general, "constant noise" (sometimes referred to as constant background or baseline) is caused by factors that are relatively stable over time, including but not limited to electroactive species that arise from generally constant (e.g., daily) metabolic processes. In contrast, "non-constant noise" (sometimes referred

to as non-constant background) is caused by transient events, such as during wound healing or in response to an illness, or due to ingestion (e.g., some drugs). In particular, noise can be caused by a variety of interfering species (constant or non-constant). Interfering species can be compounds, such as drugs that have been administered to the host, or products of various host metabolic processes. Exemplary interferents include but are not limited to a variety of drugs (e.g., acetaminophen), $H_2O_2$ from exterior sources, reactive metabolic species (e.g., reactive oxygen and nitrogen species, some hormones, etc.). In some circumstances, constant noise-causing factors can have an affect on the sensor signal similar to non-constant noise-causing factors, such as when the concentration of a constant noise-causing factor temporarily increases, such as due to temporary lack of lymph flow (see discussion of intermittent sedentary noise).

**[0553]** Noise can be difficult to remove from the sensor signal by calibration using standard calibration equations (e.g., because the background of the signal does not remain constant). Noise can significantly adversely affect the accuracy of the calibration of the analyte signal. Additionally noise, as described herein, can occur in the signal of conventional sensors with electrode configurations that are not particularly designed to measure noise substantially equally at both active and in-active electrodes (e.g., wherein the electrodes are spaced and/or non symmetrical, noise may not be equally measured and therefore not easily removed using conventional dual-electrode designs).

**[0554]** Noise can be recognized and/or analyzed in a variety of ways. In preferred embodiments, the sensor data stream is monitored, signal artifacts are detected and data processing is based at least in part on whether or not a signal artifact has been detected, such as described in U.S. Patent Publication No. US-2005-0043598-A1. Additional discussion can also be found in U.S. Patent Publication No. US-2007-0032706-A1.

Reduction of Noise

**[0555]** Noise can be recognized and substantially reduced and/or eliminated by a variety of sensor configurations and/or methods, such as by using 1) sensor configurations that block and/or remove the interferent, or that specifically detect the noise and 2) mathematical algorithms that recognize and/or remove the signal noise component. The preferred embodiments provide devices and methods for reducing and/or eliminating noise, such as by blocking interferent passage to the sensor's electroactive surfaces, diluting and/or removing interferents around the sensor and mathematically determining and eliminating the noise signal component. Those knowledgeable in the art will recognize that the various sensor structures (e.g., multiple working electrodes, membrane interference domains, etc.), bioactive agents, algorithms and the like disclosed herein can be employed in a plurality of combinations, depending upon the desired effect and the noise reduction strategy selected. In preferred embodiments, the sensor comprises at least two working electrodes (one with and one without enzyme over its electroactive surface) and an interference domain configured to substantially block interferent passage therethrough, such that at least some interferent no longer has a substantial affect on sensor measurements (e.g., at either working electrode). The term "interference domain," as used herein is a broad term, and is to be given its ordinary and customary meaning to a person of ordinary skill in the art (and it is not to be limited to a special or customized meaning), and refers without limitation to any mechanism of the membrane system configured to reduce any kind of noise or interferants, such as constant and/or non-constant noise. "Noise-reducing mechanisms" as used herein is a broad term, and is to be given its ordinary and customary meaning to a person of ordinary skill in the art (and it is not to be limited to a special or customized meaning), and refers without limitation to any sensor system component configuration that reduces and/or eliminates noise on the sensor signal. Such structural configurations include but are not limited to electrode configurations (e.g., two or more working electrodes), membrane configurations (e.g., interference domain), algorithmic configurations (e.g., signal processing to remove an identified noise component of the signal), and the like. In some embodiments, the interference domain is a component of the membrane system, such as shown in **Fig. 3C.** However, the interference domain can be disposed at any level (e.g., layer or domain) of the membrane system (e.g., more proximal or more distal to the electroactive surfaces than as shown in **Fig. 3C**). In some other embodiments, the interference domain is combined with an additional membrane domain, such as the resistance domain or the enzyme domain.

**[0556]** In another aspect, the sensor is configured to reduce noise, including non-constant non-analyte related noise with an overlapping measuring potential with the analyte. A variety of noise can occur when a sensor has been implanted in a host. Generally, implantable sensors measure a signal (e.g., counts) that generally comprises at least two components, the background signal (e.g., background noise) and the analyte signal. The background signal is composed substantially of signal contribution due to factors other than glucose (e.g., interfering species, non-reaction-related hydrogen peroxide, or other electroactive species with an oxidation/reduction potential that overlaps with the analyte or co-analyte). The analyte signal (e.g., glucose) is composed substantially of signal contribution due to the analyte. Consequently, because the signal includes these two components, a calibration is performed in order to determine the analyte (e.g., glucose) concentration by solving for the equation $y=mx+b$, where the value of b represents the background of the signal.

**[0557]** In some circumstances, the background is comprised of both constant (e.g., baseline) and non-constant (e.g., noise) factors. Generally, it is desirable to remove the background signal, to provide a more accurate analyte concentration

to the host or health care professional.

**[0558]** The term "baseline" as used herein is a broad term, and is to be given its ordinary and customary meaning to a person of ordinary skill in the art (and is not to be limited to a special or customized meaning), and refers without limitation to a substantially constant signal derived from certain electroactive compounds found in the human body that are relatively constant (e.g., baseline of the host's physiology, non-analyte related). Therefore, baseline does not significantly adversely affect the accuracy of the calibration of the analyte concentration (e.g., baseline can be relatively constantly eliminated using the equation $y=mx+b$).

**[0559]** In contrast, "noise" as used herein is a broad term, and is to be given its ordinary and customary meaning to a person of ordinary skill in the art (and is not to be limited to a special or customized meaning), and refers without limitation to a substantially intermittent signal caused by relatively non-constant factors (e.g., the presence of intermittent noise-causing compounds that have an oxidation/reduction potential that substantially overlaps the oxidation/reduction potential of the analyte or co-analyte and arise due to the host's ingestion, metabolism, wound healing, and other mechanical, chemical and/or biochemical factors, also non-analyte related). Noise can be difficult to remove from the sensor signal by calibrations using standard calibration equations (e.g., because the background of the signal does not remain constant). Noise can significantly adversely affect the accuracy of the calibration of the analyte signal. Additionally noise, as described herein, can occur in the signal of conventional sensors with electrode configurations that are not particularly designed to measure noise substantially equally at both active and in-active electrodes (e.g., wherein the electrodes are spaced and/or non symmetrical, noise may not be equally measured and therefore not easily removed using conventional dual-electrode designs).

**[0560]** There are a variety of ways noise can be recognized and/or analyzed. In preferred embodiments, the sensor data stream is monitored, signal artifacts are detected, and data processing is based at least in part on whether or not a signal artifact has been detected, such as described in U.S. Patent Publication No. US-2005-0043598-A1 and U.S. Patent Publication No. US-2007-0027370-A1.

**[0561]** Accordingly, if a sensor is designed such that the signal contribution due to baseline and noise can be removed, then more accurate analyte concentration data can be provided to the host or a healthcare professional.

**[0562]** One embodiment provides an analyte sensor (e.g., glucose sensor) configured for insertion into a host for measuring an analyte (e.g., glucose) in the host. The sensor includes a first working electrode disposed beneath an active enzymatic portion of a membrane on the sensor; a second working electrode disposed beneath an inactive- or non-enzymatic portion of the membrane on the sensor, and electronics operably connected to the first and second working electrode and configured to process the first and second signals to generate an analyte (e.g., glucose) concentration substantially without signal contribution due to non-glucose related noise artifacts.

**[0563]** In one embodiment, the sensor is configured to substantially eliminate (e.g., subtract out) noise due to mechanical factors. Mechanical factors include macro-motion of the sensor, micro-motion of the sensor, pressure on the sensor, local tissue stress, and the like. Since both working electrodes are constructed substantially symmetrically and identically, and due to the sensor's small size, the working electrodes are substantially equally affected by mechanical factors impinging upon the sensor. For example, if a build-up of noise-causing compounds occurs (e.g., due to the host pressing upon and manipulating (e.g., fiddling with) the sensor, for example) both working electrodes will measure the resulting noise to substantially the same extend, while only one working electrode (the first working electrode, for example) will also measure signal due to the analyte concentration in the host's body. The sensor then calculates the analyte signal (e.g., glucose-only signal) by removing the noise that was measured by the second working electrode from the total signal that was measured by the first working electrode.

**[0564]** Non-analyte related noise can also be caused by biochemical and/or chemical factors (e.g., compounds with electroactive acidic, amine or sulfhydryl groups, urea, lactic acid, phosphates, citrates, peroxides, amino acids (e.g., L-arginine), amino acid precursors or break-down products, nitric oxide (NO), NO-donors, NO-precursors or other electroactive species or metabolites produced during cell metabolism and/or wound healing). As with noise due to mechanical factors, noise due to biochemical/chemical factors will impinge upon the two working electrodes of the preferred embodiments (e.g., with and without active GOx) about the same extent, because of the sensor's small size and symmetrical configuration. Accordingly, the sensor electronics can use these data to calculate the glucose-only signal, as described elsewhere herein.

**[0565]** In one exemplary embodiment, the analyte sensor is a glucose sensor that measures a first signal associated with both glucose and non-glucose related electroactive compounds having a first oxidation/reduction potential. For example, the oxidation/reduction potential of the non-glucose related electroactive compounds substantially overlaps with the oxidation/reduction potential of $H_2O_2$, which is produced according to the reaction of glucose with GOx and subsequently transfers electrons to the first working electrode (e.g., **E1; Fig. 3F**). The glucose sensor also measures a second signal, which is associated with background noise of the glucose sensor. The background noise is composed of signal contribution due to noise-causing compounds (e.g., interferents), non-reaction-related hydrogen peroxide, or other electroactive species with an oxidation/reduction potential that substantially overlaps with the oxidation/reduction potential of $H_2O_2$ (the co-analyte). The first and second working electrodes integrally form at least a portion of the sensor,

such as but not limited to the *in vivo* portion of the sensor, as discussed elsewhere herein. Furthermore, the sensor has a diffusion barrier that substantially blocks (e.g., attenuates) diffusion of glucose or $H_2O_2$ between the first and second working electrodes. In various embodiments, the sensor includes a diffusion barrier configured to be physical, spatial, and/or temporal.

**[0566]** **Fig. 3F** is a schematic illustrating one embodiment of a sensor (e.g., a portion of the *in vivo* portion of the sensor, such as but not limited to the sensor electroactive surfaces) having one or more components that act as a diffusion barrier (e.g., prevent diffusion of electroactive species from one electrode to another). The first working electrode **E1** is coated with an enzyme layer **348** comprising active enzyme. For example, in a glucose sensor, the first working electrode **E1** is coated with glucose oxidase enzyme (GOx). A second working electrode **E2** is separated from the first working electrode **E1** by a diffusion barrier ***D***, such as but not limited to a physical diffusion barrier (e.g., either a reference electrode or a layer of non-conductive material/insulator). The diffusion barrier can also be spatial or temporal, as discussed elsewhere herein.

**[0567]** Glucose and oxygen diffuse into the enzyme layer **348,** where they react with GOx, to produce gluconate and $H_2O_2$. At least a portion of the $H_2O_2$ diffuses to the first working electrode **E1,** where it is electrochemically oxidized to oxygen and transfers two electrons (e.g., $2e^-$) to the first working electrode **E1,** which results in a glucose signal that is recorded by the sensor electronics (not shown). The remaining $H_2O_2$ can diffuse to other locations in the enzyme layer or out of the enzyme layer (illustrated by the wavy arrows). Without a diffusion barrier ***D***, a portion of the $H_2O_2$ can diffuse to the second working electrode **E2,** which results in an aberrant signal that can be recorded by the sensor electronics as a non-glucose related signal (e.g., background).

**[0568]** Preferred embodiments provide for a substantial diffusion barrier ***D*** between the first and second working electrodes (**E1**, **E2**) such that the $H_2O_2$ cannot substantially diffuse from the first working electrode **E1** to the second working electrode **E2.** Accordingly, the possibility of an aberrant signal produced by $H_2O_2$ from the first working electrode **E1** (at the second working electrode **E2**) is reduced or avoided.

**[0569]** In some alternative embodiments, the sensor is provided with a spatial diffusion barrier between electrodes (e.g., the working electrodes). For example, a spatial diffusion barrier can be created by separating the first and second working electrodes by a distance that is too great for the $H_2O_2$ to substantially diffuse between the working electrodes. In some embodiments, the spatial diffusion barrier is about 0.01, 0.02, 0.03, 0.04, 0.05, 0.06, 0.07, or 0.08 inches to about 0.09, 0.10, 0.11, or 0.120 inches. In other embodiments, the spatial diffusion barrier is about 0.020 inches to about 0.050 inches. Still in other embodiments, the spatial diffusion barrier is about 0.055 inches to about 0.095 inches. A reference electrode **R** (e.g., a silver or silver/silver chloride electrode) or a non-conductive material (e.g., a polymer structure or coating such as Parylene) can be configured to act as a spatial diffusion barrier.

**[0570]** In some preferred embodiment, the sensor is an indwelling sensor, such as configured for insertion into the host's circulatory system via a vein or an artery. In some exemplary embodiments, an indwelling sensor includes at least two working electrodes that are inserted into the host's blood stream through a catheter. The sensor includes at least a reference electrode that can be disposed either with the working electrodes or remotely from the working electrodes. The sensor includes a spatial, a physical, or a temporal diffusion barrier. A spatial diffusion barrier can be configured as described in US Patent Application 11/865,572, filed October 1, 2007 and entitled "DUAL ELECTRODE SYSTEM FOR A CONTINUOUS ANALYTE SENSOR ".

**[0571]** In one exemplary embodiment of an indwelling analyte sensor, such as but not limited to an intravascular glucose sensor to be used from a few hours to ten days or longer. Namely, the sensor includes two working electrodes. A first working electrode detects the glucose-related signal (due to active GOx applied to the electroactive surface) as well as non-glucose related signal. The second working electrode detects only the non-glucose related signal (because no active GOx is applied to its electroactive surface). $H_2O_2$ is produced on the first working electrode (with active GOx). If the $H_2O_2$ diffuses to the second working electrode (the no GOx electrode) an aberrant signal will be detected at this electrode, resulting in reduced sensor activity. Accordingly, it is desirable to separate the electroactive surfaces with a diffusion barrier, such as but not limited to a spatial diffusion barrier. Indwelling sensors are described in more detail in copending U.S. patent application 11/543,396 filed on October 4, 2006 and entitled "ANALYTE SENSOR".

**[0572]** To configure a spatial diffusion barrier between the working electrodes, the location of the active enzyme (e.g., GOx) is dependent upon the orientation of the sensor after insertion into the host's artery or vein. For example, in an embodiment configured for insertion in the host's blood flow (e.g., in an artery or vein), active GOx and the inactive GOX (or no GOx) would be applied to two working electrodes such that the active GOX would be downstream from the inactive GOX (e.g., relative to the direction of blood flow). Due to this configuration, $H_2O_2$ produced at plus-GOX electroactive surface would be carrier down stream (e.g., away from minus-GOX electroactive surface) and thus not affect the non-enzymatic working electrode.

**[0573]** In some embodiments, a physical diffusion barrier is provided by a physical structure, such as an electrode, insulator, and/or membrane. For example, in some embodiments, an insulator or reference electrode disposed between the working electrodes acts as a diffusion barrier. As another example, the diffusion barrier can be a bioprotective membrane (e.g., a membrane that substantially resists, attenuates or blocks the transport of a species (e.g., hydrogen

peroxide), such as a polyurethane. As yet another example, the diffusion barrier can be a resistance domain, as described in more detail elsewhere herein; namely, a semipermeable membrane that controls the flux of oxygen and an analyte (e.g., glucose) to the underlying enzyme domain. Numerous other structures and membranes can function as a physical diffusion barrier as is appreciated by one skilled in the art.

**[0574]** In other embodiments, a temporal diffusion barrier is provided (e.g., between the working electrodes). By temporal diffusion barrier is meant a period of time that substantially prevents an electroactive species (e.g., $H_2O_2$) from diffusing from a first working electrode to a second working electrode. For example, in some embodiments, the differential measurement can be obtained by switching the bias potential of each electrode between the measurement potential and a non-measurement potential. The bias potentials can be held at each respective setting (e.g., high and low bias settings) for as short as milliseconds to as long as minutes or hours. Pulsed amperometric detection (PED) is one method for quickly switching voltages, such as described in Bisenberger, M.; Brauchle, C.; Hampp, N. A triple-step potential waveform at enzyme multisensors with thick-film gold electrodes for detection of glucose and sucrose. Sensors and Actuators 1995, B, 181-189.

**[0575]** In some embodiments, bias potential settings are held long enough to allow equilibration.

**[0576]** One preferred embodiment provides a glucose sensor configured for insertion into a host for measuring glucose in the host. The sensor includes first and second working electrodes and an insulator located between the first and second working electrodes. The first working electrode is disposed beneath an active enzymatic portion of a membrane on the sensor and the second working electrode is disposed beneath an inactive- or non-enzymatic portion of the membrane on the sensor. The sensor also includes a diffusion barrier configured to substantially block (e.g., attenuate, restrict, suppress) diffusion of glucose or hydrogen peroxide between the first and second working electrodes.

**[0577]** In a further embodiment, the glucose sensor includes a reference electrode configured integrally with the first and second working electrodes. In some embodiments, the reference electrode can be located remotely from the sensor, as described elsewhere herein. In some embodiments, the surface area of the reference electrode is at least six times the surface area of the working electrodes. In some embodiments, the sensor includes a counter electrode that is integral to the sensor or is located remote from the sensor, as described elsewhere herein.

**[0578]** In a further embodiment, the glucose sensor detects a first signal associated with glucose and non-glucose related electroactive compounds having a first oxidation/reduction potential (e.g., the oxidation/reduction potential of $H_2O_2$). In some embodiments, the glucose sensor also detects a second signal is associated with background noise of the glucose sensor comprising signal contribution due to interfering species, non-reaction-related hydrogen peroxide, or other electroactive species with an oxidation/reduction potential that substantially overlaps with the oxidation/reduction potential of hydrogen peroxide; the first and second working electrodes integrally form at least a portion of the sensor; and each of the first working electrode, the second working electrode and the non-conductive material/insulator are configured provide at least two functions such as but not limited to electrical conductance, insulation, structural support, and a diffusion barrier

**[0579]** In further embodiments, the glucose sensor includes electronics operably connected to the first and second working electrodes. The electronics are configured to calculate at least one analyte sensor data point using the first and second signals described above. In still another further embodiment, the electronics are operably connected to the first and second working electrode and are configured to process the first and second signals to generate a glucose concentration substantially without signal contribution due to non-glucose noise artifacts.

Sensor Configurations for Equivalent Measurement of Noise Signals at the Two Working Electrodes

**[0580]** In dual-electrode biosensors (e.g., an analyte sensor having two working electrodes **E1, E2**), noise can be caused by a variety of sources, for example, located outside (e.g., by noise-causing species produced metabolically and/or consumed by the host) or within (e.g., crosstalk) the sensor. In some circumstances, biological and/or metabolic processes occurring in the host's body, such as in the locale of the implanted sensor, can cause noise. These metabolic processes, such as but not limited to wound healing, the body's response to illness and even daily cellular metabolic processes, can generate noise-causing metabolic species (e.g., compounds, substances) that impinge upon the sensor and cause noise on the signal. For example, some noise-causing species, the levels of which are relatively stable due to production during daily cellular metabolism, generally cause constant noise. In another example, some noise-causing species, the levels of which fluctuate due to production by intermittent metabolic process (e.g., wound healing or response to infection), generally cause non-constant noise. Noise-causing metabolic species include but are not limited to externally generated $H_2O_2$ (e.g., produced outside the sensor), compounds having electroactive acidic, amine or sulfhydryl groups, urea, lactic acid, phosphates, citrates, peroxides, amino acids (*e.g.*, L-arginine), amino acid precursors or break-down products, nitric oxide (NO), NO-donors, NO-precursors, reactive oxygen species or other electroactive species or metabolites produced during cell metabolism and/or wound healing, for example. Noise-causing species, such as drugs, vitamins and the like, can also be consumed by the host. These noise causing species include but are not limited to acetaminophen, ascorbic acid, dopamine, ephedrine, ibuprofen, L-dopa, methyldopa, salicylate, tetracycline, tolazamide,

tolbutamide and triglycerides. Further discussion of noise and its sources can be found in U.S. Patent Publication No. US-2007-0027370-A1 and U.S. Patent Publication No. US-2007-0235331-A1.

**[0581]** In dual-electrode sensors, noise can also be generated within the sensor, namely due to diffusion of a measured species (e.g., $H_2O_2$) from a first working electrode (e.g., the $H_2O_2$ is generated in an active enzymatic portion of the sensor membrane associated with the first working electrode) to a second working electrode and detection thereby (e.g., which is associated with a non-enzymatic portion of the sensor membrane). This type of noise is commonly referred to as "crosstalk." Crosstalk is undesirable as it causes sensor error, which can result in inaccurate reporting of sensor data. In conventional sensors, a common solution to the problem of crosstalk is to space the two working electrodes far enough apart that a measured species diffusing from one working electrode cannot reach the other working electrode; unfortunately, such spacing does not enable substantially equivalent measurement of noise-cause species, as discussed in more detail elsewhere herein. Unlike conventional sensors, the sensors of the preferred embodiments ensure accurate subtraction of noise signal by ensuring substantially equivalent measurement of the noise (e.g., noise component, constant and/or non-constant noise components) detected by the two working electrodes.

**[0582]** Depending upon the scale (e.g., point) of reference, noise has a dual nature. On a larger scale, with respect to the *in vivo* portion of the sensor and the surrounding tissue, noise occurs randomly (e.g., is scattered, intermittent, dispersed, unevenly distributed) in the local of an implanted sensor. Yet, on a smaller scale, such as that of a few cells (e.g., 100-300 microns), noise is a localized phenomenon because it creates hot spots of noise-causing species generation whose effects extend about a thousandths of an inch (e.g., localized nature, character). A "hot spot" of noise generation is referred to herein as a "point source." A point source (e.g., a localized hot spot for noise generation) can be a cell or a group of cells adjacent to the sensor membrane, or a noise-causing species (e.g., compound, substance, molecule) that diffused to the location of sensor implantation, such as by diffusion between cells (e.g., to the sensor). For example, in the circumstance of a single point source in contact with the sensor membrane's surface, noise is a local phenomenon, because the noise-causing species' ability to affect adjacent structures is limited by the maximum distance it can diffuse (e.g., through the membrane), which is generally very short (e.g., a few microns, such as between about 1-μm to about 500-μm). Due to the random yet localized nature of noise, the configuration of the electroactive surfaces (of the working electrodes) can substantially affect noise measurement. With respect to the configuration and arrangement (e.g., surface area) of the dual-electrode sensor's electroactive surfaces, the random yet localized nature of noise is discussed in greater detail below.

**[0583]** **Fig. 3G** is a two-dimensional schematic illustrating, on the scale of a sensor and the surrounding tissue (e.g., a generally larger scale), the random nature of noise relative to a dual-electrode sensor, in one exemplary embodiment. This figure is for illustrative purposes only, and should not be considered as a to-scale representation of a particular sensor configuration or of the events discussed herein. In the embodiment shown in **Fig. 3G,** the dual-electrode analyte sensor includes two electroactive surfaces **E1, E2** disposed beneath the sensor's membrane. While **Fig. 3G** illustrates only one dimension of the electroactive surfaces, in some embodiments, the electroactive surfaces (e.g., the surface area of each electroactive surface) can include both a length and a width. In some embodiments, the area can include additional dimensions, such as a circumference and/or a height. In some embodiments, the sensor can have a planar configuration. In some embodiments, the sensor can have a cylindrical, pyramidal, polygonal configuration. It should also be understood that the electroactive surfaces **E1, E2** are shown as boxes as a matter of illustrative convenience; however, electroactive surfaces can be thinner or thicker than illustrated in **Fig. 3G** or elsewhere herein. The membrane has a thickness ***D1*** and a surface **MS.** Depending upon the membrane configuration, fabrication methods and/or materials, ***D1*** can vary in size, from less than about 0.001, 0.002, 0.003, 0.004, 0.005, 0.006, 0.007, 0.008, 0.009, or 0.010 inches to more than about 0.011, 0.012, 0.013, 0.014, 0.015, 0.015, 0.016, 0.017, 0.018, 0.019, 0.020, 0.025, 0.03, 0.035, or 0.050 inches. In some embodiments, a preferred membrane thickness is between about 0.001, 0.0012, 0.0014, 0.0016, or 0.0018 inches to about 0.002, 0.0022, 0.0024, 0.0026, 0.0028, or 0.003 inches. Noise-causing species (represented by squiggly arrows **N2**) can be generated by and/or at point sources **N1** (e.g., noise hot spots) unevenly distributed relative the *in vivo* portion of the sensor. For example, some of the point sources **N1** (shown in **Fig. 3G**) are concentrated at one end of electroactive surface **E1,** while some are distributed more evenly across electroactive surface **E2.** In some circumstances, the point source may be one or more cells (e.g., in contact with the membrane surface **MS**) that release the noise-causing species during wound healing or another metabolic process, such as when a sensor is implanted *in vivo*. In some circumstances, the implanted sensor can be located within the diffusion distance of one or more noise-causing species produced during a nearby metabolic process. In some circumstances, the noise-causing species (e.g., a compound consumed by the host) can be carried to the local of the sensor *via* the circulatory and/or lymph system and diffuse to the sensor (e.g., between cells).

**[0584]** Random and/or unequally distributed noise can be generated in a variety of circumstances. For example, a peroxide-generating immune cell could be located adjacent to one electroactive surface but not the other. In general, a noise-causing species must be generated and/or occur close enough to the sensor membrane such that it can diffuse to (and through) the membrane, to the electroactive surfaces, and affect the sensor signal. If the noise-causing species is generated farther away from the membrane than the diffusion distance of the noise-causing species, then the noise-

causing species may be unable to reach the electroactive surfaces, and therefore may have little effect on sensor signal. For example, $H_2O_2$ (produced by metabolic process when the sensor is implanted in a host) must be generated sufficiently close to the membrane for it to diffuse to the membrane and affect sensor function. The maximum distance that the noise-causing species can diffuse (e.g., from the cell to the membrane, from one working electrode to another working electrode) and still substantially affect sensor function is referred to herein as a "diffusion distance."

**[0585]** The sensor electronics are configured to mathematically correct for noise on the sensor signal (e.g., such as by subtraction of the noise signal, applying a filter, averaging, or other calculations), such that a substantially analyte-only signal can be presented to the user. The inventors have discovered that successful mathematical correction of noise on the sensor signal can be substantially affected by the equivalence (e.g., similarity) of the noise signals detected by the two working electrodes. If the detected noise signals are substantially equivalent (e.g., similar amounts, amplitudes, levels, relatively equal), then the calculations will produce a more accurate resultant analyte signal. If, on the other hand, the detected noise signals are not substantially equal (e.g., have very different amplitudes and/or wave forms), then the calculations will have a greater degree of error. While not wishing to be bound by theory, it is believed that presentation of more accurate sensor data (e.g., to the host) will improve the host's management of his diabetes, which will prevent the immediate risks of hypoglycemia (e.g., loss of consciousness and death) and postpone and/or prevent long term diabetes complications (blindness, loss of limb, kidney dysfunction, and the like). Additionally, the increased accuracy afforded by the sensors of the preferred embodiment increases the feasibility of insulin dosing and/or an artificial pancreas system based on a continuous glucose sensor.

**[0586]** In order to compensate for the unevenly distributed nature of noise (e.g., the point sources are randomly and/or non-equally and/or non-equivalently distributed relative to the *in vivo* portion of the sensor) and thereby render the noise components equivalent, a continuous dual-electrode glucose sensor having sufficiently large electroactive surfaces, such that the noise components can be substantially equalized (e.g., made and/or become equivalent) by integration there across, is provided in one embodiment. The first working electrode includes a first electroactive surface (**E1, Fig. 3G**) disposed beneath an active enzymatic portion (e.g., plus-GOx) of the sensor's membrane, as described elsewhere herein. The first electroactive surface includes a first area (e.g., first electroactive surface area) configured to detect a first signal (e.g., including an analyte-related component and a noise component) having a first noise component related to a noise-causing species. The sensor also includes a second working electrode having a second electroactive surface (**E2, Fig. 3G**) disposed beneath an inactive-enzymatic or a non-enzymatic portion of the sensor membrane, as described elsewhere herein. For example, an inactive-enzymatic portion of the membrane can include inactivated GOx or no GOx. The second electroactive surface includes a second area (e.g., second electroactive surface area) configured to generate a second signal having a second noise component related to the noise-causing species. In preferred embodiments, the first and second areas are dimensioned (e.g., sized) to be sufficiently large such that the first and second noise components integrated there across, such that the first and second integrated noise signals (e.g., from the first and second electroactive surfaces, respectively) are substantially equivalent. In some embodiments, the first and second integrated noise signals (e.g., noise components) are within 20% of each other (e.g., plus or minus 10%). In some embodiments, the first and second electroactive surfaces are dimensioned to integrate noise caused by a plurality of local point sources that produce noise-causing species *in vivo.*

**[0587]** **Fig. 3H** is a two-dimensional schematic illustrating the localized character of noise species (represented by squiggly arrows **N2**) generated by a point source **N1,** when examined from a cellular scale, as discussed elsewhere herein. **Fig. 3H** depicts a cross-section of a sensor, in one embodiment, wherein the sensor includes two working electrodes having electroactive surfaces **E1** and **E2,** and a membrane having surface **MS** and thickness ***D1.*** Distance ***D3*** separates the electroactive surfaces; the distance between their outer edges is denoted by ***D4***. Note that dimension ***D2,*** described above, is not shown in this figure. The sensor of this exemplary embodiment can have a variety of configurations, such as but not limited to planar, cylindrical or polygonal. Accordingly, the dimensions of an electroactive surface's surface area (referred to as "area" herein) can include but is not limited to length, width, height, and/or circumference. For example, in some embodiments, the area of each electroactive surface is defined by a length and a width. In some embodiments, the area includes a length or width and a circumference. In some embodiments, the area includes length, width and height. Additionally, it is known to one skilled in the art that, in some circumstances, differences in signal amplitude and/or sensitivity (e.g., from the two working electrodes), due to differences in electrode sizes (or some differences in compositions of membranes) can be corrected (adjusted, compensated for, calibrated out) mathematically. For example, if a first electroactive surface is two times as large as the second electroactive surface, then the signal from the second electroactive surface can be multiplied by two, such that the sensitivities are substantially similar.

**[0588]** Referring now to **Fig. 3H,** in this exemplary circumstance the point source (e.g., noise hot spot) is an individual cell **N1** disposed adjacent to the membrane surface **MS** and generally above and/or over the sensor's electroactive surfaces **E1, E2.** The cell can produce noise-causing substances (e.g., **N2**) that can diffuse to and affect its local environment. In general, the ability of a noise-causing substance to affect the local environment is limited by the maximum distance the substance can diffuse (e.g., the substance's diffusion distance). In some circumstances, some of the noise-causing substances can diffuse through the sensor membrane and affect the sensor's electroactive surfaces. While not

wishing to be bound by theory, the inventors have found that in order for the two electroactive surfaces to be substantially equivalently affected by the noise **N2** from a point source, such as a cell, the electroactive surfaces must be affected by substantially the same microenvironment. In various circumstances, the electroactive surfaces will be affected by substantially the same microenvironment, if the electroactive surfaces are configured and arranged such that the electroactive surfaces are sufficiently close together and/or their external edges are sufficiently close together.

**[0589]** **Fig. 3H** shows that the sensor's electroactive surfaces **E1, E2** are separated by a distance ***D3*** and their outer edges are spaced a distance ***D4*** (e.g., in at least one dimension), in one exemplary embodiment. In this example, a point source **N1** (e.g., a cell) of noise-causing species **1006** is adjacent to the membrane's surface **MS.** If the electroactive surfaces are configured and arranged such that ***D3*** is sufficiently small, then the noise-causing species diffusing from the point source can impinge equivalently on both of the electroactive surfaces. Additionally or alternatively, if ***D4*** is sufficiently small (e.g., the electroactive surfaces are sufficiently narrow in at least one dimension), then the noise-causing species diffusing from the point source can impinge equivalently on both of the electroactive surfaces. Accordingly, in preferred embodiments, the electroactive surfaces are spaced a distance (e.g., relative to each other, ***D3***) such that the electroactive surfaces (e.g., at least a portion of each electroactive surface) detect substantially equivalent noise from a point source. In some embodiments, the electroactive surfaces are sufficiently close together (e.g., such that the noise components measured are substantially equal) when the distance between the electroactive surfaces (***D3***) is between about 0.5-times to about 10-times (or more) the membrane thickness (***D1***). In some preferred embodiments, the electroactive surfaces are sufficiently close together when ***D3*** is about 2-, 3-, 4-, 5-, 6-, 7-, 8-, or 9-times the membrane thickness. In some embodiments, ***D3*** is between about 5, 10, 15, 20, 25, 30, 35, 40, 45, or 50 microns or less to about 55, 60, 65, 70, 75, 80, 85, 90, 95, or 100 microns or more. In preferred embodiments, ***D3*** is between about 20 to about 40 microns. In some embodiments, ***D4*** is between about 25 microns or less to about 500 microns or more.

**[0590]** Depending upon the sensor's configuration, in some embodiments, ***D4*** can be the distance between the outer edges of the electroactive surfaces, or ***D4*** can be a distance equivalent to the maximum diameter of the bundles and/or twisted pair of working electrodes. For example, **Fig. 3H** illustrates a cross-section of a sensor (e.g., width and height), but doesn't illustrate any additional dimensions (e.g., length). The cross-section could be that of a planar sensor configuration, wherein the sensor also includes an additional dimension that has not been shown, such as but not limited to ***D2***. In some circumstances, the sensor can have a non-planar configuration. For example, in the embodiment shown in **Fig. 3I**, the working electrodes **E1, E2** are fabricated from two wires. Since the wires are cylindrical, the electroactive surfaces do not include outer edges. In this exemplary circumstance, ***D4*** is the total diameter of the bundled and/or twisted pair of working electrodes. In both types of sensor configurations (e.g., planar and non-planar), if ***D4*** is sufficiently small, then the two working electrodes can be equivalently affected by noise-causing species **N2** derived from a point source **N1**.

**[0591]** As described above, dual-electrode sensors can be affected by internally generated noise (e.g., generated by the sensor). The inventors have found that, in general, when ***D3*** is sized to be sufficiently small such that the electroactive surfaces are equivalently affect by noise from an adjacent point source, the electroactive surfaces are also close enough together that crosstalk (an internally generated noise) can occur. In general, crosstalk is detection of an analyte signal generated at the plus-GOx working electrode (wherein the electrode includes the membrane portion thereon) by the minus-GOx working electrode (including the No GOx membrane portion thereon). For example, when the measured species is $H_2O_2$, crosstalk occurs when the $H_2O_2$ diffuses from the plus-GOx enzyme domain to the No GOx working electrode and is detected (e.g., a signal is generated on the No GOx electrode). In general, crosstalk is undesirable as it causes sensor error. However, in order for the two working electrodes to measure equivalent noise signals from a point source **N1,** the electroactive surfaces must be spaced very close together. Accordingly, in preferred embodiments, this distance (***D3***) is less than a crosstalk diffusion distance of the measured species. In other words, ***D3*** is shorter than the diffusion distance of $H_2O_2$ (e.g., the maximum distance $H_2O_2$ can diffuse from a first electrode to a second and still cause a signal on the second electrode).

**[0592]** In conventional dual-electrode sensors, spacing the electroactive surfaces within the crosstalk diffusion distance of the measured species is generally undesirable due to increased sensor error. However, in preferred embodiments, the sensor includes a physical diffusion barrier configured to attenuate crosstalk by physically blocking (e.g., suppressing, blocking, restricting) some of the crosstalk from the active enzymatic portion of the sensor membrane to the second electroactive surface. More preferably, the physical diffusion barrier is configured and arranged to attenuate and/or physically block a substantial amount of the measurable species (e.g., $H_2O_2$) diffusing from the active enzymatic portion of the membrane to the second electroactive surface, such that there is substantially no signal associated with crosstalk measured at the second working electrode.

**[0593]** **Fig. 3I** is a schematic illustrating a perspective view of a cross-section of a dual-electrode sensor that includes a physical diffusion barrier **D**, in one exemplary embodiment. In this embodiment, wires form the working electrodes **E1, E2.** The working electrodes each include a membrane, including an electrode domain **347,** an enzyme domain **248** and a resistance domain **249.** For example, **E1** includes a first electrode domain, a first enzyme domain (Plus GOx) and a first resistance domain, and **E2** includes a second electrode domain, a second enzyme domain (No GOx) and a second

resistance domain. In this particular exemplary embodiment, the electrodes are placed together and coated with an additional resistance domain **249A** (e.g., a third resistance domain). Depending upon the circumstances, the electrodes can be placed and/or held together using a variety of methods, such as bundling, twisting, wrapping, and the like, either alone or in combination. The distances shown are as follows; a thickness of the membrane ***D1***, at least one dimension of the electroactive surface ***D2,*** a distance between the electroactive surfaces ***D3,*** and a distance between the outer edges of the electroactive surfaces ***D4.*** In the illustrated exemplary embodiment, the first and second electroactive surfaces extend about the circumferences of **E1** and **E2** (or portions thereof), respectively.

**[0594]** In preferred embodiments, the physical diffusion barrier **D** is disposed between the electroactive surfaces of working electrodes **E1** and **E2.** In some embodiments, the physical diffusion barrier is formed of one or more membrane materials, such as those used in formation of an interference domain and/or a resistance domain. Such materials include but are not limited to silicones, polyurethanes, cellulose derivatives (cellulose butyrates and cellulose acetates, and the like) and combinations thereof, as described elsewhere herein. In some embodiments, the physical diffusion barrier includes one or more membrane domains. For example, in the exemplary embodiment of **Fig. 3I,** the physical diffusion barrier is a discontinuous portion of the membrane (e.g., separate, distinct or discontinuous membrane structures) disposed between the first and second electroactive surfaces, and can include one or more membrane portion(s) within distance ***D3*** (e.g., interference and/or resistance domains). For example, in some embodiments, $H_2O_2$ diffusing from the Plus GOX working electrode to the No GOx working electrode must pass through two "sensor membranes" such as the first and second resistance domains disposed on **E1** and **E2** respectively, and optionally electrode, interference and/or enzyme domains disposed on **E2.** In some embodiments, the physical diffusion barrier includes first and second barrier layers formed independently on the first and second electrodes. In some embodiments the barrier layer is the resistance domain **349.** In still other embodiments, the physical diffusion barrier can be a continuous membrane (and/or membrane domain(s)) disposed between the electroactive surfaces. In some embodiments, the physical diffusion barrier attenuates (e.g., suppresses, blocks, prevents) diffusion of the $H_2O_2$ (e.g., crosstalk) by at least 2-fold. In preferred embodiments, crosstalk is attenuated at least 5-fold. In a more preferred embodiment, crosstalk is attenuated at least 10-fold. In some embodiments, the physical diffusion barrier attenuates crosstalk at least about 50%. In a further embodiment, the physical diffusion barrier is configured and arranged to physically block an amount of the measured species diffusing from the active enzymatic portion of the membrane to the second electroactive surface, such that there is substantially no signal associated with crosstalk measured at the second working electrode.

**[0595]** In some embodiments, a dual-electrode sensor having a physical barrier layer can be fabricated by initially preparing (e.g., fabricating, building) the first and second working electrodes **E1, E2** independently (e.g., separately from each other), followed by joining and/or grouping and/or bundling the working electrodes and optionally applying one or more additional membrane domains fabrication. In this exemplary embodiment, to the first working electrode **E1,** an optional electrode domain **347,** an enzyme domain **348** (e.g., plus-GOx), and at least one layer of the resistance domain material **349** (e.g., first resistance domain) are sequentially applied. Similarly, to the second working electrode **E2,** an optional electrode domain **347,** an enzyme domain **348** (e.g., no-GOx), and at least one layer of the resistance domain material **349** (e.g., second resistance domain) are sequentially applied. The working electrodes are then held together, such as but not limited to by bundling and/or twisting them together, wrapping a material around them, or by any other method known in the art. In this embodiment, the physical diffusion barrier **D** includes a discontinuous portion of a membrane (e.g., the initial layers of the resistance domain material applied independently to the two working electrodes) disposed between the first and second electroactive surfaces.

**[0596]** In an alternative exemplary sensor embodiment, the sensor includes working electrodes (including electroactive surfaces) disposed on a planar substrate and/or surface. The electroactive surfaces can be spaced a distance ***D3*** that is sufficiently close together that the electroactive surfaces are equivalently affected by an adjacent noise hot spot (e.g., point source). In this configuration, ***D3*** is also sufficiently small that crosstalk can occur between the Plus GOx working electrode (wherein the term "electrode" includes the membrane disposed thereon, for the purposes of this example) and the No GOx working electrode. However, in preferred embodiments, crosstalk is substantially attenuated by a physical diffusion barrier disposed between the working electrodes. Namely, the electrode domains (if present) and enzyme domains can be separately applied to the working electrodes and/or electroactive surfaces; followed by application of a continuous resistance domain applied thereon, such that a portion the resistance domain is deposited between the working electrodes. For example, a portion of resistance domain deposited on a planar substrate and between working electrodes can attenuate diffusion of the measured species (e.g., $H_2O_2$) from **E1** to **E2,** such that the noise measured on **E1** and **E2** is equivalent.

**[0597]** In the context of glucose sensors, one skilled in the art recognizes that equivalent noise signals can have different amplitudes, but equivalent signal patterns (e.g., rises, falls, trends and the like) such that a noise component can be subtracted out (as described elsewhere herein) while compensating for any difference in signal amplitude (e.g., sensitivity of the first and second working electrodes), as described elsewhere herein. In some circumstances, the membrane portions associated with the working electrodes (e.g., of a dual-electrode sensor) can possess different sensitivities (e.g., signal sensitivities), such that the amplitudes of the noise components measured by the working

electrodes are not equivalent. In some circumstances, the areas of the electroactive surfaces may be different sizes, which can also result in non-equivalent signal amplitudes, differences in measured baselines and/or sensitivities between the first and second working electrodes. While such differences in signal baseline and/or sensitivity can be corrected mathematically (e.g., by mathematical filters), mathematical correction of noise, in general, is improved when the signal sensitivities of the first and second working electrodes are closer. Accordingly, in a preferred embodiment, an additional resistance domain **349A** (e.g., applied continuously over the discontinuous resistance domains **349** described elsewhere herein) is provided, such that the signal sensitivities are equivalent. In the exemplary embodiment shown in **Fig. 3l,** the signal sensitivities are substantially equalized on a sensor including the combination of discontinuous resistance domains (e.g., resistance domains **349,** applied independently to **E1** and **E2**) and a continuous resistance domain **349A** (e.g., applied over and/or adjacent to the discontinuous resistance domains). In other words, the noise signals detected on both **E1** and **E2** will have substantially the same amplitude (e.g., intensity, amount), as described with reference to **Example 7,** below. In a preferred embodiment, the sensitivities (of the working electrodes) are within 40% of each other (e.g., plus or minus 20%). In a preferred embodiment, the sensitivities (of the working electrodes) are within 20% of each other (e.g., plus or minus 10%). In a more preferred embodiment, the sensitivities (of the working electrodes) are within 10% of each other (e.g., plus or minus 5%).

**[0598]** In an alternative embodiment, the sensor electrodes can be disposed on a planar, cylindrical, pyramidal or otherwise shaped support. For example, the sensor's first and second working electrodes can be conductive traces deposited, such as by screen printing, sputtering or other thin film techniques known in the art, on a planar substrate. In this alternative embodiment, a physical diffusion barrier can be formed by layers of resistance domain material deposited separately (e.g., discontinuously) on each working electrode and/or between the electrodes, for example.

**[0599]** In the exemplary embodiments described above, diffusion of the $H_2O_2$ from the first working electrode **E1** to the electroactive surface of the second working electrode **E2** is first attenuated by the resistance domain **349** disposed over the first working electrode **E1** (an independently formed first barrier layer), and then again by the resistance domain **349** disposed over the second working electrode **E2** (an independently formed second barrier layer), such that only insubstantial amounts of $H_2O_2$ can reach the electroactive surface of the second working electrode. In preferred embodiments, the first and second resistance domains are configured and arranged to reduce diffusion of the measurable species (e.g., $H_2O_2$) from the first electroactive surface to the second electroactive surface by at least 2-fold. In more preferred embodiments, the physical diffusion barrier is configured and arranged to reduce diffusion of the measurable species by at least 10-fold. In some embodiments, the physical diffusion barrier is configured and arranged to reduce diffusion of the measurable species by at least 3-, 4-, 5-, 6-, 7-, 8-, or 9-fold. In some embodiments, the physical diffusion barrier is configured and arranged to reduce diffusion of the measurable species by at least 20-, 30-, 40- or 50- fold, or more. In some embodiments, the sensor's working electrodes E1, E2 are by an insulator, which insulates the working electrodes from each other. In some embodiments, the insulator is at least a portion of the sensor membrane.

**[0600]** In some embodiments, a continuous glucose sensor configured for insertion into a host and detecting glucose in the host is provided. In general, the sensor includes first and second working electrodes, wherein each working electrode includes an electroactive surface (each including an area) disposed beneath a sensor membrane. The first electroactive surface (e.g., of the first working electrode) is disposed beneath an active enzymatic portion (plus-GOx) of the membrane while the second electroactive surface (of the second working electrode) is disposed beneath a non-enzymatic (no-GOx) portion of the membrane. The non-enzymatic portion of the membrane can include inactivated enzyme and/or no enzyme. Additionally, each working electrode is configured to generate a signal having a noise component related to a noise-causing species. In some circumstances, the noise-causing species is non-constant and related to a biological process. Preferably, the first and second areas are sufficiently large such that the noise components (e.g., first and second noise components detected by the first and second working electrodes) are substantially equivalent. In some embodiments, the first and second areas are each greater than the sum of the diameters of about 10 average human cells, in at least one dimension. In some embodiments, the first and second areas are each greater than about 500$\mu$m, in at least one dimension. Preferably, the first and second areas (e.g., of the electroactive surfaces) are configured and arranged such that the signals caused by a plurality of local point sources (that produce noise-causing species when implanted in a host) can be integrated along each area (e.g., each area independently from the other). In some further embodiments, the first and second areas are configured and arranged to integrate signals detected about a circumference of the sensor. Preferably, the first and second electroactive surfaces are spaced a distance that is less than a crosstalk diffusion distance of a measured species, such as $H_2O_2$ produced in the active enzymatic portion of the membrane. In some embodiments, the sensor includes a physical diffusion barrier configured and arranged to physically block some crosstalk from the active enzymatic portion of the membrane to the second electroactive surface. In some further embodiments, the physical diffusion barrier is configured and arranged to physically block a substantial amount of the measurable species diffusing from the active enzymatic portion of the membrane to the second electroactive surface (e.g., crosstalk), such that there is substantially no signal associated with crosstalk measured at the second working electrode. In some embodiments, the physical diffusion barrier is a discontinuous portion of the membrane disposed between the first and second electroactive surfaces. In some further embodiments, the physical diffusion barrier includes

a first barrier layer formed on the first working electrode and a second barrier layer formed on the second working electrode, wherein the first and second barrier layers are independently formed (e.g., formed separately on the two electroactive surfaces). In some further embodiments, the physical diffusion barrier includes a first resistance domain formed on the first working electrode and a second resistance domain formed on the second working electrode, and wherein the first and second resistance domains are configured and arranged to reduce diffusion of the measurable species (e.g., crosstalk) from the active enzymatic portion of the sensor to the second electroactive surface by at least 2-fold. In a preferred embodiment, the physical diffusion barrier can reduce the diffusion of the measurable species (e.g., crosstalk) by at least 10-fold.

**[0601]** In some embodiments, the continuous glucose sensor includes first and second working electrodes, each working electrode including an electroactive surface (each including an area) disposed beneath a sensor membrane. As described elsewhere herein, the first electroactive surface is disposed beneath an active enzymatic portion of the membrane and the second electroactive surface is disposed beneath a non-enzymatic portion of the membrane. Preferably, the sensor includes a physical diffusion barrier, and the first and second electroactive surfaces are disposed sufficiently close together that the first and second noise components (detected by the first and second working electrodes) are substantially equivalent. In some embodiments, the distance between the first and second electroactive surfaces is less than about twice the thickness of the membrane. In some embodiments, the first and second electroactive surfaces are spaced a distance that is less than or equal to about a crosstalk diffusion distance of a measurable species, such as the $H_2O_2$ produced in the active enzymatic portion of the sensor membrane. In some embodiments, the physical diffusion barrier is configured and arranged to physically block some diffusion of the measurable species from the active enzymatic portion of the membrane to the second electroactive surface (e.g., crosstalk). In preferred embodiments, the physical diffusion barrier blocks a substantial amount of the measurable species, such that there is substantially no signal associated with crosstalk measured at the second working electrode. In some embodiments, the physical diffusion barrier is a discontinuous portion of the membrane disposed between the first and second electroactive surfaces. In some embodiments, the physical diffusion barrier is a first barrier layer formed on the first electrode and a second barrier layer formed on the second electrode, wherein the first and second barrier layers are independently formed. In some embodiments, the physical diffusion barrier includes a first resistance domain formed on the first electrode and a second resistance domain formed on the second electrode. Preferably, the first and second resistance domains reduce diffusion of the measurable species (e.g., crosstalk) by at least 2-fold. In more preferred embodiments, the diffusion of the measurable species is reduced by at least 10-fold. In some embodiments, the membrane is an insulator that insulates the first working electrode from the second working electrodes. In some further embodiments, the first and second areas are sufficiently large that the first and second noise components are substantially equivalent.

Sensor Electronics

**[0602]** The analyte sensor system has electronics, also referred to as a "computer system" that can include hardware, firmware, and/or software that enable measurement and processing of data associated with analyte levels in the host. In one exemplary embodiment, the electronics include a potentiostat, a power source for providing power to the sensor, and other components useful for signal processing. In another exemplary embodiment, the electronics include an RF module for transmitting data from sensor electronics to a receiver remote from the sensor. In another exemplary embodiment, the sensor electronics are wired to a receiver, which records the data and optionally transmits the data to a remote location, such as but not limited to a nurse's station, for tracking the host's progress and to alarm the staff is a hypoglycemic episode occurs. In another exemplary embodiment, the sensor electronics include a processor module for processing sensor data, as described elsewhere herein. In some exemplary embodiments, the sensor electronics include a receiving module for receiving sensor signals, such as but not limited to from the working electrode(s), and/or externally provided reference data points. In some embodiments, the processor module can include the receiving module. The processor module and the receiving module can be located together and/or in any combination of sensor electronics local to and/or remote from the sensor.

**[0603]** Various components of the electronics of the sensor system can be disposed on or proximal to the analyte sensor, such as but not limited to disposed on the fluid coupler **20** of the system, such as the embodiment shown in Fig. 1A. In another embodiment, wherein the sensor is integrally formed on the catheter (e.g., see Fig. 2A) and the electronics are disposed on or proximal to the connector **218.** In some embodiments, only a portion of the electronics (e.g., the potentiostat) is disposed on the device (e.g., proximal to the sensor), while the remaining electronics are disposed remotely from the device, such as on a stand or by the bedside. In a further embodiment, a portion of the electronics can be disposed in a central location, such as a nurse's station.

**[0604]** In additional embodiments, some or all of the electronics can be in wired or wireless communication with the sensor and/or other portions of the electronics. For example, a potentiostat disposed on the device can be wired to the remaining electronics (e.g., a processor, a recorder, a transmitter, a receiver, etc.), which reside on the bedside. In another example, some portion of the electronics is wirelessly connected to another portion of the electronics, such as

by infrared (IR) or RF. In one embodiment, a potentiostat resides on the fluid coupler and is connected to a receiver by RF; accordingly, a battery, RF transmitter, and/or other minimally necessary electronics are provided with the fluid coupler and the receiver includes an RF receiver.

**[0605]** Preferably, the potentiostat is operably connected to the electrode(s) (such as described above), which biases the sensor to enable measurement of a current signal indicative of the analyte concentration in the host (also referred to as the analog portion). In some embodiments, the potentiostat includes a resistor that translates the current into voltage. In some alternative embodiments, a current to frequency converter is provided that is configured to continuously integrate the measured current, for example, using a charge counting device.

**[0606]** In some embodiments, the electronics include an A/D converter that digitizes the analog signal into a digital signal, also referred to as "counts" for processing. Accordingly, the resulting raw data stream in counts, also referred to as raw sensor data, is directly related to the current measured by the potentiostat.

**[0607]** Typically, the electronics include a processor module that includes the central control unit that controls the processing of the sensor system. In some embodiments, the processor module includes a microprocessor, however a computer system other than a microprocessor can be used to process data as described herein, for example an ASIC can be used for some or all of the sensor's central processing. The processor typically provides semi-permanent storage of data, for example, storing data such as sensor identifier (ID) and programming to process data streams (for example, programming for data smoothing and/or replacement of signal artifacts such as is described in U.S. Patent Publication No. US-2005-0043598-A1). The processor additionally can be used for the system's cache memory, for example for temporarily storing recent sensor data. In some embodiments, the processor module comprises memory storage components such as ROM, RAM, dynamic-RAM, static-RAM, non-static RAM, EEPROM, rewritable ROMs, flash memory, and the like.

**[0608]** In some embodiments, the processor module comprises a digital filter, for example, an infinite impulse response (IIR) or finite impulse response (FIR) filter, configured to smooth the raw data stream from the A/D converter. Generally, digital filters are programmed to filter data sampled at a predetermined time interval (also referred to as a sample rate). In some embodiments, wherein the potentiostat is configured to measure the analyte at discrete time intervals, these time intervals determine the sample rate of the digital filter. In some alternative embodiments, wherein the potentiostat is configured to continuously measure the analyte, for example, using a current-to-frequency converter as described above, the processor module can be programmed to request a digital value from the A/D converter at a predetermined time interval, also referred to as the acquisition time. In these alternative embodiments, the values obtained by the processor are advantageously averaged over the acquisition time due the continuity of the current measurement. Accordingly, the acquisition time determines the sample rate of the digital filter. In preferred embodiments, the processor module is configured with a programmable acquisition time, namely, the predetermined time interval for requesting the digital value from the A/D converter is programmable by a user within the digital circuitry of the processor module. An acquisition time of from about 2 seconds to about 512 seconds is preferred; however any acquisition time can be programmed into the processor module. A programmable acquisition time is advantageous in optimizing noise filtration, time lag, and processing/battery power.

**[0609]** In some embodiments, the processor module is configured to build the data packet for transmission to an outside source, for example, an RF transmission to a receiver. Generally, the data packet comprises a plurality of bits that can include a preamble, a unique identifier identifying the electronics unit, the receiver, or both, (e.g., sensor ID code), data (e.g., raw data, filtered data, and/or an integrated value) and/or error detection or correction. Preferably, the data (transmission) packet has a length of from about 8 bits to about 128 bits, preferably about 48 bits; however, larger or smaller packets can be desirable in certain embodiments. The processor module can be configured to transmit any combination of raw and/or filtered data. In one exemplary embodiment, the transmission packet contains a fixed preamble, a unique ID of the electronics unit, a single five-minute average (e.g., integrated) sensor data value, and a cyclic redundancy code (CRC).

**[0610]** In some embodiments, the processor module further comprises a transmitter portion that determines the transmission interval of the sensor data to a receiver, and the like. In some embodiments, the transmitter portion, which determines the interval of transmission, is configured to be programmable. In one such embodiment, a coefficient can be chosen (*e.g.*, a number of from about 1 to about 100, or more), wherein the coefficient is multiplied by the acquisition time (or sampling rate), such as described above, to define the transmission interval of the data packet. Thus, in some embodiments, the transmission interval is programmable from about 2 seconds to about 850 minutes, more preferably from about 30 second to about 5 minutes; however, any transmission interval can be programmable or programmed into the processor module. However, a variety of alternative systems and methods for providing a programmable transmission interval can also be employed. By providing a programmable transmission interval, data transmission can be customized to meet a variety of design criteria (*e.g.*, reduced battery consumption, timeliness of reporting sensor values, etc.)

**[0611]** In some embodiments, the processor further performs the processing, such as storing data, analyzing data streams, calibrating analyte sensor data, estimating analyte values, comparing estimated analyte values with time cor-

responding measured analyte values, analyzing a variation of estimated analyte values, downloading data, and controlling the user interface by providing analyte values, prompts, messages, warnings, alarms, and the like. In such cases, the processor includes hardware and software that performs the processing described herein, for example flash memory provides permanent or semi-permanent storage of data, storing data such as sensor ID, receiver ID, and programming to process data streams (for example, programming for performing estimation and other algorithms described elsewhere herein) and random access memory (RAM) stores the system's cache memory and is helpful in data processing. Alternatively, some portion of the data processing (such as described with reference to the processor elsewhere herein) can be accomplished at another (e.g., remote) processor and can be configured to be in wired or wireless connection therewith.

**[0612]** In some embodiments, an output module, which is integral with and/or operatively connected with the processor, includes programming for generating output based on the data stream received from the sensor system and it's processing incurred in the processor. In some embodiments, output is generated via a user interface.

**[0613]** In some embodiments, a user interface is provided integral with (e.g., on the patient inserted medical device), proximal to (e.g., a receiver near the medical device including bedside or on a stand), or remote from the sensor electronics (e.g., at a central station such as a nurse's station), wherein the user interface comprises a keyboard, speaker, vibrator, backlight, liquid crystal display (LCD) screen, and one or more buttons. The components that comprise the user interface include controls to allow interaction of the user with the sensor system. The keyboard can allow, for example, input of user information, such as mealtime, exercise, insulin administration, customized therapy recommendations, and reference analyte values. The speaker can produce, for example, audible signals or alerts for conditions such as present and/or estimated hyperglycemic or hypoglycemic conditions. The vibrator can provide, for example, tactile signals or alerts for reasons such as described with reference to the speaker, above. The backlight can be provided, for example, to aid a user in reading the LCD in low light conditions. The LCD can be provided, for example, to provide the user with visual data output, such as is described in U.S. Patent Publication No. US-2005-0203360-A1. In some embodiments, the LCD is a touch-activated screen, enabling each selection by a user, for example, from a menu on the screen. The buttons can provide for toggle, menu selection, option selection, mode selection, and reset, for example. In some alternative embodiments, a microphone can be provided to allow for voice-activated control.

**[0614]** In some embodiments, prompts or messages can be displayed on the user interface to convey information to the user, such as reference outlier values, requests for reference analyte values, therapy recommendations, deviation of the measured analyte values from the estimated analyte values, and the like. Additionally, prompts can be displayed to guide the user through calibration or trouble-shooting of the calibration.

**[0615]** Additionally, data output from the output module can provide wired or wireless, one- or two-way communication between the user interface and an external device. The external device can be any device that wherein interfaces or communicates with the user interface. In some embodiments, the external device is a computer, and the system is able to download historical data for retrospective analysis by the patient or physician, for example. In some embodiments, the external device is a modem or other telecommunications station, and the system is able to send alerts, warnings, emergency messages, and the like, via telecommunication lines to another party, such as a doctor or family member. In some embodiments, the external device is an insulin pen, and the system is able to communicate therapy recommendations, such as insulin amount and time to the insulin pen. In some embodiments, the external device is an insulin pump, and the system is able to communicate therapy recommendations, such as insulin amount and time to the insulin pump. The external device can include other technology or medical devices, for example pacemakers, implanted analyte sensor patches, other infusion devices, telemetry devices, and the like.

**[0616]** The user interface, including keyboard, buttons, a microphone (not shown), and optionally the external device, can be configured to allow input of data. Data input can be helpful in obtaining information about the patient (for example, meal time, insulin administration, and the like), receiving instructions from a physician (for example, customized therapy recommendations, targets, and the like), and downloading software updates, for example. Keyboard, buttons, touch-screen, and microphone are all examples of mechanisms by which a user can input data directly into the receiver. A server, personal computer, personal digital assistant, insulin pump, and insulin pen are examples of external devices that can provide useful information to the receiver. Other devices internal or external to the sensor that measure other aspects of a patient's body (for example, temperature sensor, accelerometer, heart rate monitor, oxygen monitor, and the like) can be used to provide input helpful in data processing. In one embodiment, the user interface can prompt the patient to select an activity most closely related to their present activity, such as medication taken, surgical procedures, and the like, which can be helpful in linking to an individual's physiological patterns, or other data processing. In another embodiment, a temperature sensor and/or heart rate monitor can provide information helpful in linking activity, metabolism, and glucose excursions of an individual. While a few examples of data input have been provided here, a variety of information can be input, which can be helpful in data processing.

Algorithms

**[0617]** In some embodiments, calibration of an analyte sensor can be required, which includes data processing that

converts sensor data signal into an estimated analyte measurement that is meaningful to a user. In general, the sensor system has a computer system (e.g., within the electronics) that receives sensor data (*e.g.*, a data stream), including one or more time-spaced sensor data points, measured by the sensor. The sensor data point(s) can be smoothed (filtered) in certain embodiments using a filter, for example, a finite impulse response (FIR) or infinite impulse response (IIR) filter. During the initialization of the sensor, prior to initial calibration, the system can receive and store uncalibrated sensor data, however it can be configured to not display any data to the user until initial calibration and, optionally, stabilization of the sensor has been established. In some embodiments, the data stream can be evaluated to determine sensor break-in (equilibration of the sensor *in vitro* or *in vivo*).

**[0618]** In some embodiments, the system is configured to receive reference data from a reference analyte monitor, including one or more reference data points, also referred to as calibration information in some embodiments. The monitor can be of any suitable configuration. For example, in one embodiment, the reference analyte points can comprise results from a self-monitored blood analyte test (e.g., from a finger stick test, YSI, Beckman Glucose Analyzer, and the like), such as those described in U.S. Pat. Nos. 6,045,567; 6,156,051; 6,197,040; 6,284,125; 6,413,410; and 6,733,655. In one such embodiment, the user can administer a self-monitored blood analyte test to obtain an analyte value (e.g., point) using any suitable analyte sensor, and then enter the numeric analyte value into the computer system. In another such embodiment, a self-monitored blood analyte test comprises a wired or wireless connection to the computer system so that the user simply initiates a connection between the two devices, and the reference analyte data is passed or downloaded between the self-monitored blood analyte test and the system. In yet another such embodiment, the self-monitored analyte test is integral with the receiver so that the user simply provides a blood sample to the receiver, and the receiver runs the analyte test to determine a reference analyte value.

**[0619]** In some alternative embodiments, the reference data is based on sensor data from another substantially continuous analyte sensor such as described herein, or another type of suitable continuous analyte sensor. In an embodiment employing a series of two or more continuous sensors, the sensors can be employed so that they provide sensor data in discrete or overlapping periods. In such embodiments, the sensor data from one continuous sensor can be used to calibrate another continuous sensor, or be used to confirm the validity of a subsequently employed continuous sensor.

**[0620]** In some embodiments, the sensor system is coupled to a blood analysis device that periodically or intermittently collects a sample of the host's blood (e.g., through the sensor system) and measures the host's glucose concentration. In some embodiments, the blood analysis device collects a blood sample from the host about every 30 minutes, every hour, or every few hours (e.g., 2, 3, 4, 5, 6, 8, 9 or 10 hours or longer). In other embodiments, the blood analysis device can be activated manually (e.g., by a healthcare worker) to collect and analyze a blood sample from the host. The glucose concentration data generated by the blood analysis device can be used by the sensor system for calibration data. In some embodiments, the sensor system can electronically receive (either wired or wirelessly) these calibration data (from the blood analysis device). In other embodiments, these calibration data can be entered into the sensor system (e.g., sensor system electronics) by hand (e.g., manually entered by a healthcare worker).

**[0621]** In some embodiments, the sensor system is provided with one or more calibration solutions (e.g., glucose solutions). In some embodiments, the sensor is shipped in a calibration solution (e.g., soaked). The sensor is activated to calibrate itself (using the calibration solution in which it was shipped) before insertion into the host. In some embodiments, the sensor is shipped (e.g., soaked or dry) with one or more vials of calibration solution. The sensor can be soaked (e.g., sequentially) in the vial(s) of calibration solution; calibration data points collected and the sensor calibrated using those calibration points, before inserting the sensor into the host.

**[0622]** In one exemplary embodiment, the sensor is a glucose sensor, and it is shipped soaking in a sterile 50 mg/dl glucose solution with two accompanying calibration solutions (e.g., 100 mg/dl and 200 mg/dl sterile glucose solutions). Prior to insertion into the host, calibration data points are collected with the sensor in the 50 mg/dl, 100 mg/dl and 200 mg/dl glucose solutions respectively. The sensor system can be calibrated using the collected calibration data points (e.g., using regression as described in more detail elsewhere herein). In an alternative exemplary embodiment, the sensor is shipped dry (e.g., not soaking in a solution or buffer) with at least one calibration solution, for calibrating the sensor prior to insertion into the host. In some embodiments, a hand held glucose monitor (e.g., SMBG device described herein) can test the calibration solutions to generate calibration data points, which are transferred electronically or manually to the sensor system for calibration.

**[0623]** In some embodiments, a data matching module, also referred to as the processor module, matches reference data (*e.g.*, one or more reference analyte data points) with substantially time corresponding sensor data (*e.g.*, one or more sensor data points) to provide one or more matched data pairs. One reference data point can be matched to one time corresponding sensor data point to form a matched data pair. Alternatively, a plurality of reference data points can be averaged (*e.g.*, equally or non-equally weighted average, mean-value, median, and the like) and matched to one time corresponding sensor data point to form a matched data pair, one reference data point can be matched to a plurality of time corresponding sensor data points averaged to form a matched data pair, or a plurality of reference data points can be averaged and matched to a plurality of time corresponding sensor data points averaged to form a matched data pair.

**[0624]** In some embodiments, a calibration set module, also referred to as the calibration module or processor module,

forms an initial calibration set from a set of one or more matched data pairs, which are used to determine the relationship between the reference analyte data and the sensor analyte data. The matched data pairs, which make up the initial calibration set, can be selected according to predetermined criteria. The criteria for the initial calibration set can be the same as, or different from, the criteria for the updated calibration sets. In certain embodiments, the number (n) of data pair(s) selected for the initial calibration set is one. In other embodiments, n data pairs are selected for the initial calibration set wherein n is a function of the frequency of the received reference data points. In various embodiments, two data pairs make up the initial calibration set or six data pairs make up the initial calibration set. In an embodiment wherein a substantially continuous analyte sensor provides reference data, numerous data points are used to provide reference data from more than 6 data pairs (*e.g.*, dozens or even hundreds of data pairs). In one exemplary embodiment, a substantially continuous analyte sensor provides 288 reference data points per day (every five minutes for twenty-four hours), thereby providing an opportunity for a matched data pair 288 times per day, for example. While specific numbers of matched data pairs are referred to in the preferred embodiments, any suitable number of matched data pairs per a given time period can be employed.

**[0625]** In some embodiments, a conversion function module, also referred to as the conversion module or processor module, uses the calibration set to create a conversion function. The conversion function substantially defines the relationship between the reference analyte data and the analyte sensor data.

**[0626]** A variety of known methods can be used with the preferred embodiments to create the conversion function from the calibration set. In one embodiment, wherein a plurality of matched data points form the calibration set, a linear least squares regression is used to calculate the conversion function; for example, this regression calculates a slope and an offset using the equation $y=\boldsymbol{mx}+b$. A variety of regression or other conversion schemes can be implemented herein.

**[0627]** In some alternative embodiments, the sensor is a dual-electrode system. In one such dual-electrode system, a first electrode functions as a hydrogen peroxide sensor including a membrane system containing glucose-oxidase disposed thereon, which operates as described herein. A second electrode is a hydrogen peroxide sensor that is configured similar to the first electrode, but with a modified membrane system (with the enzyme domain removed, for example). This second electrode provides a signal composed mostly of the baseline signal, *b*.

**[0628]** In some dual-electrode systems, the baseline signal is (electronically or digitally) subtracted from the glucose signal to obtain a glucose signal substantially without baseline. Accordingly, calibration of the resultant difference signal can be performed by solving the equation $y = \boldsymbol{mx}$ with a single paired measurement. Calibration of the implanted sensor in this alternative embodiment can be made less dependent on the values/range of the paired measurements, less sensitive to error in manual blood glucose measurements, and can facilitate the sensor's use as a primary source of glucose information for the user. U.S. Patent Publication No. US-2005-0143635-A1 describes systems and methods for subtracting the baseline from a sensor signal.

**[0629]** In some alternative dual-electrode system embodiments, the analyte sensor is configured to transmit signals obtained from each electrode separately (*e.g.*, without subtraction of the baseline signal). In this way, the receiver can process these signals to determine additional information about the sensor and/or analyte concentration. For example, by comparing the signals from the first and second electrodes, changes in baseline and/or sensitivity can be detected and/or measured and used to update calibration (e.g., without the use of a reference analyte value). In one such example, by monitoring the corresponding first and second signals over time, an amount of signal contributed by baseline can be measured. In another such example, by comparing fluctuations in the correlating signals over time, changes in sensitivity can be detected and/or measured.

**[0630]** In some alternative embodiments, a regression equation $y=\boldsymbol{mx}+b$ is used to calculate the conversion function; however, prior information can be provided for $\boldsymbol{m}$ and/or *b*, thereby enabling calibration to occur with fewer paired measurements. In one calibration technique, prior information (*e.g.*, obtained from *in vivo* or *in vitro* tests) determines a sensitivity of the sensor and/or the baseline signal of the sensor by analyzing sensor data from measurements taken by the sensor (*e.g.*, prior to inserting the sensor). For example, if there exists a predictive relationship between *in vitro* sensor parameters and *in vivo* parameters, then this information can be used by the calibration procedure. For example, if a predictive relationship exists between *in vitro* sensitivity and *in vivo* sensitivity, $\boldsymbol{m} \approx f(\boldsymbol{m}_{in\ vitro})$, then the predicted $\boldsymbol{m}$ can be used, along with a single matched pair, to solve for $b$ ($b = y - \boldsymbol{mx}$). If, in addition, *b* can be assumed = 0, for example with a dual-electrode configuration that enables subtraction of the baseline from the signal such as described above, then both $\boldsymbol{m}$ and *b* are known *a priori*, matched pairs are not needed for calibration, and the sensor can be completely calibrated *e.g.* without the need for reference analyte values (*e.g.* values obtained after implantation *in vivo*.)

**[0631]** In another alternative embodiment, prior information can be provided to guide or validate the baseline (*b*) and/or sensitivity ($\boldsymbol{m}$) determined from the regression analysis. In this embodiment, boundaries can be set for the regression line that defines the conversion function such that working sensors are calibrated accurately and easily (with two points), and non-working sensors are prevented from being calibrated. If the boundaries are drawn too tightly, a working sensor may not enter into calibration. Likewise, if the boundaries are drawn too loosely, the scheme can result in inaccurate calibration or can permit non-working sensors to enter into calibration. For example, subsequent to performing regression, the resulting slope and/or baseline are tested to determine whether they fall within a predetermined acceptable threshold

(boundaries). These predetermined acceptable boundaries can be obtained from *in vivo* or *in vitro* tests (*e.g.*, by a retrospective analysis of sensor sensitivities and/or baselines collected from a set of sensors/patients, assuming that the set is representative of future data).

**[0632]** In some alternative embodiments, the sensor system does not require initial and/or update calibration by the host; in these alternative embodiments, also referred to as "zero-point calibration" embodiments, use of the sensor system without requiring a reference analyte measurement for initial and/or update calibration is enabled. In general, the systems and methods of the preferred embodiments provide for stable and repeatable sensor manufacture, particularly when tightly controlled manufacturing processes are utilized. Namely, a batch of sensors of the preferred embodiments can be designed with substantially the same baseline (*b*) and/or sensitivity (***m***) (+/- 10%) when tested *in vitro.* Additionally, the sensor of the preferred embodiments can be designed for repeatable ***m*** and *b in vivo.* Thus, an initial calibration factor (conversion function) can be programmed into the sensor (sensor electronics and/or receiver electronics) that enables conversion of raw sensor data into calibrated sensor data solely using information obtained prior to implantation (namely, initial calibration does not require a reference analyte value). Additionally, to obviate the need for recalibration (update calibration) during the life of the sensor, the sensor is designed to minimize drift of the sensitivity and/or baseline over time *in vivo.* Accordingly, the preferred embodiments can be manufactured for zero point calibration.

**[0633]** In some embodiments, a sensor data transformation module, also referred to as the calibration module, conversion module, or processor module, uses the conversion function to transform sensor data into substantially real-time analyte value estimates, also referred to as calibrated data, or converted sensor data, as sensor data is continuously (or intermittently) received from the sensor. For example, the sensor data, which can be provided to the receiver in "counts," is translated in to estimate analyte value(s) in mg/dL. In other words, the offset value at any given point in time can be subtracted from the raw value (*e.g.*, in counts) and divided by the slope to obtain the estimate analyte value:

$$mg/dL = \frac{(rawvalue - offset)}{slope}$$

**[0634]** In some embodiments, an output module provides output to the user via the user interface. The output is representative of the estimated analyte value, which is determined by converting the sensor data into a meaningful analyte value. User output can be in the form of a numeric estimated analyte value, an indication of directional trend of analyte concentration, and/or a graphical representation of the estimated analyte data over a period of time, for example. Other representations of the estimated analyte values are also possible, for example audio and tactile.

**[0635]** In some embodiments, annotations are provided on the graph; for example, bitmap images are displayed thereon, which represent events experienced by the host. For example, information about meals, medications, insulin, exercise, sensor insertion, sleep, and the like, can be obtained by the receiver (by user input or receipt of a transmission from another device) and displayed on the graphical representation of the host's glucose over time. It is believed that illustrating a host's life events matched with a host's glucose concentration over time can be helpful in educating the host to his or her metabolic response to the various events.

**[0636]** In yet another alternative embodiment, the sensor utilizes one or more additional electrodes to measure an additional analyte. Such measurements can provide a baseline or sensitivity measurement for use in calibrating the sensor. Furthermore, baseline and/or sensitivity measurements can be used to trigger events such as digital filtering of data or suspending display of data, all of which are described in more detail in U.S. Patent Publication No. US-2005-0143635-A1.

**[0637]** In one exemplary embodiment, the sensor can be calibrated by a calibration solution. For example, after the sensor system has been inserted into the host, a calibration solution can be injected so as to pass across the electroactive surface of the analyte-measuring electrode and the sensor calibrated thereby. For example, the saline drip can be changed to a known IV glucose or dextrose solution (e.g., D50 - a 50% dextrose solution, or D5W - a 5% dextrose solution). In one embodiment, a known volume of D5W is infused into the host at a known rate over a predetermined period of time (e.g., 5, 10, 15 or 20 minutes, or for shorter or longer periods). During and/or after the period of infusion, the sensor measures the signal at the analyte-measuring working electrode. The system, knowing the specifications of the infused calibration solution (also referred to as a calibration information in some embodiments), can calibrate the signal to obtain host's glucose concentration as is appreciated by one skilled in the art. In a further embodiment, two or more glucose or dextrose solutions can be infused, with a corresponding signal being measured during each infusion, to provide additional data for sensor calibration. Calibration can be performed after the sensor has first been inserted into the host, after a break-in time, at two or more different levels (high/low), regularly, intermittently, in response to sensor drift/shift, automatically or any other time when calibration is required. In some alternative embodiments, calibration can be determined during sensor break-in, such as described in more detail elsewhere herein.

**[0638]** In some circumstances, catheters are flushed with saline. For example, the analyte sensor system of the

preferred embodiments can be flushed with saline prior to application of control solutions, after which a predetermined amount of glucose solution is flushed by the sensor, as described above, and the sensor is calibrated there from.

**[0639]** In still another embodiment, a blood sample can be withdrawn from an artery or vein, and used to calibrate the sensor, for example, by using a hand-held glucose meter, by an automatic extracorporeal glucose sensor such as but not limited to in conjunction with an automated bedside clinical chemistry device, or by sending the blood sample to the clinical laboratory for glucose analysis, after which the data is input (e.g., into the electronics associated with the sensor system).

**[0640]** In some embodiments, the sensor can be calibrated (and/or re-calibrated) during use (after initial calibration), for example, by withdrawing one or more blood samples (also referred to as calibration information in some embodiments), through the catheter (see Figs. 1 and 2) and used for calibration of the sensor, such as by measuring the glucose concentration of the blood sample with an additional system, such as but not limited to a hand-held glucose meter, optical methods or additional electrochemical methods. Blood samples can be withdrawn manually or automatically; additionally or alternatively, blood samples are withdrawn at regular intervals or at selected times, for example, using an extracorporeal blood analysis device as described herein.

**[0641]** In another embodiment of sensor calibration (and/or re-calibration) during use, a calibration solution (e.g., 40 mg/dL equivalent glucose, D540 or D5W) can be flushed through or by the sensor to enable calibration of the sensor (e.g., at one time, intermittently, or continuously), such as described in more detail above. In these embodiments, calibration solution can be flushed manually or automatically through the system; additionally or alternatively, calibration solution can be flushed at regular intervals or at selected times. In one exemplary embodiment, the system can be provided with a dual lumen, one for saline and another for the control solution. Additionally, the system is configured to automatically switch from the saline to control solution and perform the real-time system calibration, and then switch back to the saline solution.

Calibration Systems and Methods for Dual-Electrode Sensors

**[0642]** As described herein, continuous analyte sensors define a relationship between a sensor-generated signal and a reference measurement that is meaningful to a user (for example, blood glucose in mg/dL). This defined relationship must be monitored to ensure that the continuous analyte sensor maintains a substantially accurate calibration and thereby continually provides meaningful values to a user. Unfortunately, both sensitivity ***m*** and baseline *b* changes can occur during *in vivo* sensor use, which requires calibration updates (e.g., recalibration). Generally, any physical properties of the sensor or the fluid surrounding the sensor can influence diffusion or transport of molecules through the membrane, and thereby produce fluctuations in sensitivity and/or baseline, which in turn affect the sensor's calibration. These physical properties include, but are not limited to, blockage of sensor surface area due to cells and/or blood clotting at the membrane, biofouling, blood flow/sheer rate, blood pH, temperature, hematocrit, interfering drugs in the host's system, certain metabolic processes, disrupted host electrolyte balance due to disease and/or trauma, thickness and/or components of the sensor's membrane system, and the like.

**[0643]** In one aspect of the preferred embodiments, systems and methods are provided for measuring changes in sensitivity ***m***, also referred to as changes in solute transport or membrane changes, of an analyte sensor implanted in a host over a time period. Preferably, the sensitivity measurement is a signal obtained by measuring a constant analyte other than the analyte being measured by the analyte sensor. For example, in a glucose sensor, a non-glucose constant analyte is measured, wherein the signal is measured beneath the membrane system on the glucose sensor. While not wishing to be bound by theory, it is believed that by monitoring the sensitivity ***m*** over a time period, a change associated with solute transport through the membrane system (e.g., diffusion there through) can be measured and used as an indication of a sensitivity change in the analyte measurement. In other words, a membrane monitor is provided, which is capable of monitoring changes in the membrane surrounding an implantable device, thereby enabling the measurement of sensitivity changes of an analyte sensor over time.

**[0644]** In some embodiments, the analyte sensor is provided with an auxiliary electrode (e.g., a second working electrode) configured as a transport-measuring electrode disposed beneath the membrane system. The transport-measuring electrode can be configured to measure any of a number of substantially constant analytes or factors, such that a change measured by the transport-measuring electrode can be used to indicate a change in solute (for example, glucose) transport through the membrane system. Some examples of substantially constant analytes or factors that can be measured include, but are not limited to, oxygen, carboxylic acids (such as urea), amino acids, hydrogen, pH, chloride, baseline, or the like. Thus, the transport-measuring electrode provides an independent measure of changes in solute transport to the membrane, and thus sensitivity changes over time.

**[0645]** In some embodiments, the transport-measuring electrode measures analytes similar to the analyte being measured by the analyte sensor. For example, in some embodiments of a glucose sensor, water soluble analytes are believed to better represent the changes in sensitivity to glucose over time than non-water soluble analytes (due to the water-solubility of glucose), however relevant information may be ascertained from a variety of molecules. Although some

specific examples are described herein, one skilled in the art appreciates a variety of implementations of sensitivity measurements that can be used as to qualify or quantify solute transport through the membrane of the analyte sensor.

**[0646]** In one embodiment of a glucose sensor, the transport-measuring electrode is configured to measure urea, which is a water-soluble constant analyte that is known to react directly or indirectly at a hydrogen peroxide sensing electrode (similar to the working electrode of the glucose sensor example described in more detail above). In one exemplary implementation wherein urea is directly measured by the transport-measuring electrode, the glucose sensor comprises a membrane system as described in more detail above, however, does not include an active interference domain or active enzyme directly above the transport-measuring electrode, thereby allowing the urea to pass through the membrane system to the electroactive surface for measurement thereon. In one alternative exemplary implementation wherein urea is indirectly measured by the transport-measuring electrode, the glucose sensor comprises a membrane system as described in more detail above, and further includes an active uricase oxidase domain located directly above the transport-measuring electrode, thereby allowing the urea to react at the enzyme and produce hydrogen peroxide, which can be measured at the electroactive surface thereon.

**[0647]** In some embodiments, the change in sensitivity ***m*** is measured by measuring a change in oxygen concentration and to indicate when recalibration of the system may be advantageous. In one alternative embodiment, oxygen is measured using pulsed amperometric detection on the glucose-measuring working electrode (eliminating the need for a separate auxiliary electrode), such as by switching the applied potential from +0.6mV to -0.6mV. In another embodiment, the auxiliary electrode is configured as an oxygen-measuring electrode. In some embodiments, a third electrode can be configured as an oxygen-measuring electrode. In another embodiment, an oxygen sensor (not shown) is added to the glucose sensor, as is appreciated by one skilled in the art, eliminating the need for an auxiliary electrode.

**[0648]** In some embodiments, sensitivity changes in an intravascular dual-electrode continuous analyte sensor can be provided *via* a reference sensor, such as an oxygen sensor, as described in the section entitled "Optical Detection." In some embodiments, auto-calibration (e.g., without a manual, external (to the system) reference value) is enabled by exposing the dual-electrode sensor and the reference sensor simultaneously to a reference/calibration solution, whereby reference data is provided for calibration of the sensor data. Advantageously, a dual-electrode continuous analyte sensor is configured to measure baseline *b,* and changes in sensitivity ***m*** are measured by exposure of the dual-electrode sensor to the reference/calibration solution. In some embodiments, the system is configured for "on demand" auto-calibration, such as *via* configuring the system such that a user can initiate (e.g., command) auto-calibration *via* a user interface (e.g., *via* selection from a menu, pressing a pre-programmed button and the like).

**[0649]** In some alternative embodiments, sensitivity changes can be used to update calibration. For example, the measured change in transport can be used to update the sensitivity ***m*** in the calibration equation. While not wishing to be bound by theory, it is believed that in some embodiments, the sensitivity ***m*** of the calibration of the glucose sensor is substantially proportional to the change in solute transport measured by the transport-measuring electrode.

**[0650]** It should be appreciated by one skilled in the art that in some embodiments, the implementation of sensitivity measurements of the preferred embodiments typically necessitate an addition to, or modification of, the existing electronics (for example, potentiostat configuration or settings) of the glucose sensor and/or receiver.

**[0651]** In some embodiments, the signal from the oxygen measuring electrode may be digitally low-pass filtered (for example, with a passband of $0\text{-}10^{-5}$ Hz, dc-24 hour cycle lengths) to remove transient fluctuations in oxygen, due to local ischemia, postural effects, periods of apnea, or the like. Since oxygen delivery to tissues is held in tight homeostatic control, this filtered oxygen signal should oscillate about a relatively constant. In the interstitial fluid, it is thought that the levels are about equivalent with venous blood (40 mmHg). Once implanted, changes in the mean of the oxygen signal (for example, > 5%) may be indicative of change in transport through the membrane (change in sensor sensitivity and/or baseline due to changes in solute transport) and the need for system recalibration.

**[0652]** The oxygen signal may also be used in its unfiltered or a minimally filtered form to detect or predict oxygen deprivation-induced artifact in the glucose signal, and to control display of data to the user, or the method for smoothing, digital filtering, or otherwise replacement of glucose signal artifact. In some embodiments, the oxygen sensor may be implemented in conjunction with any signal artifact detection or prediction that may be performed on the counter electrode or working electrode voltage signals of the electrode system. U.S. Patent Publication No. US-2005-0043598-A1 describes some methods of signal artifact detection and replacement that may be useful such as described herein.

**[0653]** Preferably, the transport-measuring electrode is located within the same local environment as the electrode system associated with the measurement of glucose, such that the transport properties at the transport-measuring electrode are substantially similar to the transport properties at the glucose-measuring electrode.

**[0654]** In a second aspect the preferred embodiments, systems and methods are provided for measuring changes baseline, namely non-glucose related electroactive compounds in the host. Preferably the auxiliary working electrode (e.g., second working electrode, non-enzymatic working electrode) is configured to measure the baseline of the analyte sensor over time. In some embodiments, the glucose-measuring working electrode (e.g., first working electrode) is a hydrogen peroxide sensor coupled to a membrane system containing an active enzyme located above the electrode. In some embodiments, the auxiliary working electrode (e.g., second working electrode) is another hydrogen peroxide

sensor that is configured similar to the glucose-measuring working electrode however a portion of the membrane system above the base-measuring electrode does not have active enzyme therein, such as described in more detail with reference to **Fig. 3D.** The auxiliary working electrode provides a signal substantially comprising the baseline signal, *b*, which can be (for example, electronically or digitally) subtracted from the glucose signal obtained from the glucose-measuring working electrode to obtain the signal contribution due to glucose only according to the following equation:

$$\text{Signal}_{\text{glucose only}} = \text{Signal}_{\text{glucose-measuring working electrode}} - \text{Signal}_{\text{baseline-measuring working electrode}}$$

**[0655]** In some embodiments, electronic subtraction of the baseline signal from the glucose signal can be performed in the hardware of the sensor, for example using a differential amplifier. In some alternative embodiments, digital subtraction of the baseline signal from the glucose signal can be performed in the software or hardware of the sensor or an associated receiver, for example in the microprocessor.

**[0656]** One aspect the preferred embodiments provides for a simplified calibration technique, wherein the variability of the baseline has been eliminated (namely, subtracted). Namely, calibration of the resultant differential signal ($\text{Signal}_{\text{glucose only}}$) can be performed with a single matched data pair by solving the following equation:

$$y = mx$$

**[0657]** While not wishing to be bound by theory, it is believed that by calibrating using this simplified technique, the sensor is made less dependent on the range of values of the matched data pairs, which can be sensitive to human error in manual blood glucose measurements, for example. Additionally, by subtracting the baseline at the sensor (rather than solving for the baseline *b* as in conventional calibration schemes), accuracy of the sensor may increase by altering control of this variable (baseline *b*) from the user to the sensor. It is additionally believed that variability introduced by sensor calibration may be reduced.

**[0658]** In some embodiments, the glucose-measuring working electrode (e.g., first working electrode) is a hydrogen peroxide sensor coupled to a membrane system containing an active enzyme located above the electrode, such as described in more detail above; however the baseline signal is not subtracted from the glucose signal for calibration of the sensor. Rather, multiple matched data pairs are obtained in order to calibrate the sensor (for example using $y = \boldsymbol{mx} + b$) in a conventional manner, and the auxiliary/second working electrode is used as an indicator of baseline shifts in the sensor signal. Namely, the auxiliary/second working electrode is monitored for changes above a certain threshold. When a significant change is detected, the system can trigger a request (for example, from the patient or caregiver) for a new reference glucose value (for example, SMBG), which can be used to recalibrate the sensor. By using the auxiliary/second working electrode signal as an indicator of baseline shifts, recalibration requiring user interaction (namely, new reference glucose values) can be minimized due to timeliness and appropriateness of the requests. In some embodiments, the sensor is re-calibrated responsive to a baseline shifts exceeding a preselected threshold value. In some embodiments, the sensor is calibrated repeatedly at a frequency responsive to the rate-of-change of the baseline.

**[0659]** In yet another alternative embodiment, the electrode system of the preferred embodiments is employed as described above, including determining the differential signal of glucose less baseline current in order to calibrate using the simplified equation ($y = mx$), and the auxiliary/second working electrode is further utilized as an indicator of baseline shifts in the sensor signal. While not wishing to be bound by theory, it is believed that shifts in baseline may also correlate and/or be related to changes in the sensitivity $\boldsymbol{m}$ of the glucose signal. Consequently, a shift in baseline may be indicative of a change in sensitivity $\boldsymbol{m}$. Therefore, the auxiliary working electrode is monitored for changes above a certain threshold. When a significant change is detected, the system can trigger a request (for example, from the patient or caregiver) for a new reference glucose value (for example, SMBG), which can be used to recalibrate the sensor. By using the auxiliary (second) signal as an indicator of possible sensitivity changes, recalibration requiring user interaction (new reference glucose values) can be minimized due to timeliness and appropriateness of the requests.

**[0660]** In yet another alternative embodiment, wherein a dual-electrode analyte system is use, the baseline signal is (electronically or digitally) subtracted from the glucose + baseline signal to obtain a glucose signal substantially without baseline. Accordingly, calibration of the resultant difference signal can be performed by solving the equation $y = \boldsymbol{mx}$ using a sensitivity measurement ($\boldsymbol{m}$), which can be obtained from 1) the measured reference (calibrant) solution, b) by pairing a reference analyte signal with a reference analyte value (internal reference sensor example) and/or 3) with a sensitivity measurement obtained *a priori* (e.g., during sensor manufacture, such as but not limited to by testing in an isotonic solution). Accordingly, calibration of the implanted sensor in this embodiment can be less sensitive or insensitive to user error associated with providing external reference measurements, and can facilitate the sensor's use as a primary

source of glucose information for the user. However, a single external reference value (e.g., from an external test device such as SMBG and/or YSI testing of a host blood sample) can be used to calibrate the sensor in some embodiments. U.S. Patent Publication No. US-2005-0143635-A1 describes systems and methods for subtracting the baseline from a sensor signal.

**[0661]** It is noted that, in some embodiments, infrequent new matching data pairs (e.g., auto-calibration, on demand calibration) may be useful over time to recalibrate the sensor because the sensitivity ***m*** of the sensor may change over time (for example, due to maturation of the membrane that may increase or decrease the glucose and/or oxygen availability to the sensor). However, the baseline shifts that have conventionally required numerous and/or regular blood glucose reference measurements for updating calibration (for example, due to interfering species, metabolism changes, or the like) can be consistently and accurately eliminated using the systems and methods of the preferred embodiments, allowing reduced interaction from the patient (for example, requesting less frequent reference glucose values such as daily or even as infrequently as monthly).

**[0662]** An additional advantage of the sensor of the preferred embodiments includes providing a method for eliminating signal effects of interfering species, which have conventionally been problematic in electrochemical glucose sensors. Namely, electrochemical sensors are subject to electrochemical reaction not only with the hydrogen peroxide (or other analyte to be measured), but additionally may react with other electroactive species that are not intentionally being measured (for example, interfering species), which cause a change in signal strength due to this interference. In other words, interfering species are compounds with an oxidation or reduction potential that overlap with the analyte being measured. Interfering species such as acetaminophen, ascorbate, and urate, are notorious in the art of glucose sensors for producing inaccurate signal strength when they are not properly controlled. Some glucose sensors utilize a membrane system that blocks at least some interfering species, such as ascorbate and urate. Unfortunately, it is difficult to find membranes that are satisfactory or reliable in use, especially *in vivo,* which effectively block all interferants and/or interfering species (for example, see U.S. Patent No. 4,776,944, U.S. Patent No. 5,356,786, U.S. Patent No. 5,593,852, U.S Patent No. 5,776,324, and U.S. Patent No. 6,356,776).

**[0663]** The preferred embodiments are particularly advantageous in their inherent ability to eliminate the erroneous transient and non-transient signal effects normally caused by interfering species. For example, if an interferant such as acetaminophen is ingested by a host implanted with a conventional implantable electrochemical glucose sensor (namely, one without means for eliminating acetaminophen), a transient non-glucose related increase in signal output would occur. However, by utilizing the electrode system of the preferred embodiments, both working electrodes respond with substantially equivalent increased current generation due to oxidation of the acetaminophen, which would be eliminated by subtraction of the auxiliary electrode signal from the glucose-measuring electrode signal.

**[0664]** In summary, the system and methods of the preferred embodiments simplify the computation processes of calibration, decreases the susceptibility introduced by user error in calibration, and eliminates the effects of interfering species. Accordingly, the sensor requires less interaction by the patient (for example, less frequent calibration), increases patient convenience (for example, few reference glucose values), and improves accuracy (via simple and reliable calibration).

**[0665]** In another aspect of the preferred embodiments, the analyte sensor is configured to measure any combination of changes in baseline and/or in sensitivity, simultaneously and/or iteratively, using any of the above-described systems and methods. While not wishing to be bound by theory, the preferred embodiments provide for improved calibration of the sensor, increased patient convenience through less frequent patient interaction with the sensor, less dependence on the values/range of the paired measurements, less sensitivity to error normally found in manual reference glucose measurements, adaptation to the maturation of the membrane over time, elimination of erroneous signal due to non-constant analyte-related signal so interfering species, and/or self diagnosis of the calibration for more intelligent recalibration of the sensor.

**[0666]** Preferably, a dual-electrode sensor's working electrodes **E1, E2** should function identically, with identical sensitivity and/or baseline measurements. However, in some circumstances, small differences in the portions of the membrane at the first and second working electrodes can result in slight differences in membrane sensitivities (***m***) and/or the baselines (*b*) signal associated with the working electrodes. While not wishing to be bound by theory, it is believed that such small differences can arise during manufacture, for example, when **E1** and **E2** have one or more separate manufacturing steps. Additionally, the compositions of the enzyme domains can be slightly different. For example, an **E1** enzyme domain can be somewhat more hydrophilic than an **E2** enzyme domain manufactured without any added enzyme. In some circumstances, certain characteristics of the blood samples (e.g., pH, certain medicaments in the host's circulatory system, temperature, $pO_2$) to which the sensor is exposed can, amplify the effects of these differences. In some circumstances, these small differences (between **E1** and **E2**) can contribute to sensor inaccuracies. In preferred embodiments, such sensor inaccuracies can be avoided by use of a scaling factor (*k*) that is calculated by evaluating the signal response at each of the working electrodes.

**[0667]** In one preferred embodiment, a scaling factor (*k*) is calculated by evaluating the signal response after at point at which substantially all of the analyte present in a blood sample should have been used up. **Fig. 3J** is a graph that

illustrates exemplary data collected upon exposure of a dual-electrode continuous analyte sensor to a blood sample. The Y-axis represents the signal generated and the X-axis represents time. The top graph is data generated by the plus-enzyme working electrode (e.g., first working electrode, **E1**). The bottom graph is time-corresponding data generated by the minus-enzyme working electrode (e.g., non-enzymatic, second working electrode, **E2**). In general, when the sensor is exposed to a blood sample, the **E1** signal will increase **1302** until substantially all of the available analyte is used up (e.g., at **t1**) by the enzyme at the plus-enzyme working electrode. In general, as with most enzymes, when all of the substrate (e.g., analyte) is used up, the signal should plateau (e.g., line **1306**). However, in spite of the lack of substrate (e.g., analyte) for the enzyme, the signal actually continues to increase a small amount, as is shown by line **1308.** While not wishing to be bound by theory, it is believed that the signal increase **1310** is due to signals caused by non-analyte-related electroactive species, which diffuse through the membrane more slowly than the analyte. This signal increase (line **1304**) is also observed on the non-enzymatic working electrode (e.g., **E2**). Accordingly, the signal response for E2, during the time period of t1 to t2, is the difference between lines **1305** and **1304.** The scaling factor (*k*, also referred to as the "buildup ratio") can be determined by evaluating the signal response at the two working electrodes (e.g., between $\mathbf{t_1}$ and $\mathbf{t_2}$) using the formula:

$$k^{Sensor} = \frac{Signal_response_{Enzyme}}{Signal_response_{NoEnzyme}}$$

**[0668]** Accordingly, for the exemplary data shown in **Fig. 3J,** the scaling factor (*k*) is equal to the ratio of the plus-enzyme signal response **1310** to the minus-enzyme signal response **1312.** In a related embodiment, a scaling factor can be calculated by evaluating data generated at the early portion of the curve, soon after the switch from a wash/reference solution (e.g., saline, glucose, etc.) to blood, such as between about 2 second and about 2 minutes after the switch from blood to non-bodily fluid. The scaling factor can then be used to adjust the data (e.g., calibrate) for differences in membrane sensitivities, thereby providing increased sensor accuracy. In some embodiments, the dual-electrode sensor is a glucose sensor. However, dual-electrode sensors can be configured to detect other analytes as described elsewhere herein.

**[0669]** In some embodiments, the scaling factor can be determined by evaluating the signal responses of the first and second working electrodes **E1, E2** during exposure of the sensor to a non-bodily fluid (e.g., saline, reference/calibration solution, hydration fluid, wash fluid, nutritional fluid, medicament, etc.). As described elsewhere herein, an intravascularly implanted dual-electrode analyte sensor can be washed and/or calibrated between exposures to blood samples. To measure the signal response at the working electrodes, the non-bodily fluid can be held stagnant (e.g., substantially not moving) for a period of time (e.g., from **t1** to **t2**). Signal responses at the working electrodes, during the time period, can then be evaluated.

**[0670]** **Fig. 3K** is a graph that illustrates exemplary data received upon exposure of a dual-electrode sensor to a non-bodily fluid. The Y-axis represents signal generated at the working electrodes and the X-axis represents time. The top graph is data generated by the plus-enzyme working electrode (e.g., first working electrode, **E1**). The bottom graph is time-corresponding data generated by the minus-enzyme working electrode (e.g., non-enzymatic, second working electrode, **E2**). Starting at t=0, the flow of fluid over the dual-electrode continuous analyte sensor is stopped, such that the fluid is substantially stagnant (e.g., not moving). At the first working electrode **E1** (with enzyme), an increasing signal is generated, as is represented by line **1314.** An increasing signal is also generated at the second working electrode **E2** (no enzyme), as is illustrated by line **1316.** The **E1** signal response (**1318**) is the difference between lines **1315** and **1314** at **t2.** Similarly, the **E2** signal response (**1320**) is the difference between lines **1317** and **1316** at **t2.** While not wishing to be bound by theory, it is believed that the observed signal response is due to diffusion of non-analyte-related electroactive species through the plus and minus enzyme membrane portions, which are then detected at the working electrodes. The sensor scaling factor can be calculated using the equation described above.

**[0671]** In some embodiments, to improve sensor accuracy, a scaling factor can be calculated by comparing the signal response of a test sensor ($k^{TestSensor}$) to the signal response of a "perfect sensor" ($k^{PerfectSensor}$), using the following formula:

$$ScalingFactor = \frac{k^{TestSensor}}{k^{PerfectSensor}}$$

**[0672]** The signal response ($k^{PerfectSensor}$) for a perfect dual-electrode sensor (can be determined empirically by testing

a plurality of sensors in the laboratory, such as by using methods known in the art.

**[0673]** As described in more detail elsewhere herein, useful information (e.g., sensitivity and/or scaling factor) can be extrapolated from periods in which the signal is transient, for example, during sensor break-in and/or during a period of signal artifact (e.g., noise). In some embodiments, the scaling factor is determined during electrochemical break-in of the sensor. Additionally or alternatively, the scaling factor is determined during a period of signal artifact, for example, wherein the flow of fluid across the sensor manipulated (e.g., disrupted) intentionally and/or accidentally (and detected). In one exemplary embodiment, the flow control device of the preferred embodiment is configured to jitter, reciprocate and/or dither in such a way so as to more effectively wash the sensor; in this exemplary embodiment, signal artifact is induced on the signal by the induced flow turbulence, which can be used to obtain useful information by transient analysis of the signal.

**[0674]** Due to the kinetics of the signal during these transient events, a noise amplitude can be determined for each of the first and second working electrodes of a dual electrode system, and the noise amplitude compared to obtain a scaling factor. In one embodiment, the scaling factor is determined by using a residual analysis, wherein filtered (e.g., smoothed) data is compared to raw data (e.g., in local and/or remote electronics) to obtain a signal residual. In one such embodiment, a signal residual is calculated as the difference between the filtered data and the raw data. For example, at one time point (or one time period that is represented by a single raw value and single filtered value), the filtered data can be measured at 50,000 counts and the raw data can be measured at 55,500 counts, which would result in a signal residual of 5,500 counts. In some embodiments, the residuals provide the noise amplitude information, which can be compared to obtain a scaling factor. However, in some embodiments, a stream of residuals (e.g., individual time points during a kinetic period of the signal) for each of the first and second working electrodes are averaged (e.g., using a moving average, or the like), and compared, to provide noise amplitude information for each of the first and second working electrodes, which can be used to define a scaling factor.

**[0675]** In some embodiments, the manufacturer determines a baseline (e.g., $b_{offset}$) and/or a scaling factor prior to sensor use in a host, such as but not limited to testing in one or more reference solutions, such as but not limited to an isotonic solution. The prospectively determined baseline (e.g., $b_{offset}$) and/or scaling factor can be included with the sensor provided to the user, such as by providing a calibration code that can be entered (e.g., manually) into the system electronics, that can be automatically detected by the system electronics upon sensor coupling thereto (e.g., via a detectable memory), and the like, similar to the manufacturer-provided calibration codes for glucose test strips.

**[0676]** As a non-limiting example, in preferred embodiments, a system for measuring an analyte, wherein differences in first and second working electrodes is accounted for, is provided. In this embodiment, the system includes a continuous analyte sensor, a vascular access device, a receiving module, and a processing module. The continuous analyte sensor is configured for exposure to a host's circulatory system *in vivo,* such as *via* fluidly coupling with a vascular access device in fluid contact with the host's circulatory system. The continuous analyte sensor includes first and second working electrodes **E1, E2**. The first working electrode **E1** is disposed beneath an enzymatic portion of a membrane system, wherein the enzymatic portion includes an enzyme for detecting the analyte. For example, if the analyte is glucose, the enzymatic portion includes GOX. If the analyte is cholesterol, the enzyme is a cholesterol-metabolizing enzyme. The second working electrode **E2** is disposed beneath a non-enzymatic portion of the membrane system, which includes either no enzyme or an inactive form of the enzyme. For example, the enzyme can be inactivated by a variety of methods, such as denaturing by heating, UV exposure, treatment with a protease or a denaturing chemical, and the like. In some embodiments, the enzyme layer of the membrane system over **E2** (e.g., the electroactive surface) includes another protein, such as BSA or ovalbumin, which is not involved in the metabolism of the analyte.

**[0677]** The system includes a receiving module configured to receive the signals from the working electrodes (e.g., a first signal from **E1** and a second signal from **E2**). As described elsewhere herein, the first signal is associated with both the analyte and non-analyte related electroactive compounds; the second signal is associated with non-analyte related electroactive compounds. The non-analyte related compounds have an oxidation/reduction potential that substantially overlaps with the analyte's oxidation/reduction potential. Accordingly, if the dual-electrode sensor is configured to detect glucose, **E1** detects a signal having components associated with glucose and non-glucose species that have oxidation/reduction potentials that substantially overlap with the oxidation and/or reduction potential of glucose (sometimes referred to herein as a first oxidation/reduction potential), and **E2** detects a signal related to the non-glucose species that have oxidation/reduction potentials that substantially overlap with the oxidation and/or reduction potential of glucose.

**[0678]** The system includes a processor module configured to process the first and second signals and to estimate a scaling factor. As described herein, the scaling factor defines a relationship between the first and second working electrodes (e.g., associated with the measured baseline of each first and second working electrodes). Preferably, the processor module processes the first and second signals using the scaling factor, to thereby obtain a signal (e.g., a glucose value) substantially without contribution due to non-analyte related electroactive compounds. For example, wherein the equation $b=kz$ defines the relationship (scaling factor (*k)*) between the baseline (*b)* of the first (enzymatic) working electrode and the baseline *z* of the second (non-enzymatic) working electrode. A calibration equation (y=mx+b) can be modified to include the scaling factor to calibrate the sensor (y-kz=mx).

**[0679]** Preferably, the system includes a flow control device is configured to meter a flow of a fluid through the vascular access device. In some embodiments, the fluid is a bodily fluid and the flow control device is configured to withdraw a sample of bodily fluid (e.g., blood) from the host such that the sensor is contacted with the bodily fluid. In a further embodiment, the processor module is configured to compare steady-state information of the first signal and steady-state information of the second signal. In some embodiments, the fluid is a non-bodily fluid and the flow control device is configured to hold the non-bodily fluid substantially stagnant during a time period, as described herein. In preferred embodiments, the processor module is configured to compare a signal increase on each of the first and second working electrodes during the time period during which the non-bodily fluid is held stagnant, as described herein.

**[0680]** In preferred embodiments, a method for processing sensor data from a dual-electrode continuous analyte sensor, including estimating a scaling factor (*k*), is provided. The dual-electrode sensor, as described herein, is configured for *in vivo* exposure to a host's circulatory system. The dual-electrode continuous analyte sensor is applied to the host, such as *via* fluidly coupling the sensor to a fluid flow device. In some embodiments, the sensor is configured for insertion into a catheter or is a part of the catheter, as described above. In some embodiments, the sensor is part of a connecting device, such as a Leur lock, which is fluidly coupled to a catheter at a first end and to the rest of the fluid flow device (e.g., *via* IV tubing), such that blood samples can be withdrawn and contacted with the sensor. After the sensor has been applied to the host, signals from the working electrodes can be received, as described elsewhere herein. A scaling factor, which defines a relationship between a the first and second working electrodes (e.g., a baseline associated therewith), can be estimated from the received signals, and then the scaling factor can be used to process the signals and thereby to obtain a signal substantially without contribution due to non-analyte related electroactive compounds. In some embodiments, the scaling factor is determined while contacting the sensor with a bodily fluid (see above), such as by comparing steady-state information of the first signal and steady-state information of the second signal. In some embodiments, the scaling factor is determined while contacting the sensor with a substantially stagnant non-bodily fluid, such as by comparing a signal increase on each of the working electrodes during exposure to the substantially stagnant non-bodily fluid.

**[0681]** While not wishing to be bound by theory, it is believed that determination of a scaling factor as described herein provides a number of advantages. First, sample collection/testing is alternated with calibration and/or washing, sensor calibration is continuous and biofouling is substantially reduced. Since there is little biofouling, the sensor functions more rapidly (e.g., $T_{90}$ is reached more rapidly). Since the sensor is continuously calibrated during use (e.g., such that background is removed) the user receives more accurate glucose information/values to be used in making therapeutic decisions. Thus, it is substantially easier and safer for the host to maintain tight control over his glucose levels, which can result in a better quality of life and reduced long-term diabetic complications.

**[0682]** In many circumstances, glucose sensors can be calibrated using data provided either by an analyte testing device separate from the continuous analyte sensor system. For example, points for a continuous glucose sensor, one or more reference data may be provided by testing a blood sample with a hand-held glucose meter or with an YSI glucose test device. However, in some preferred embodiments, a system, including a continuous analyte sensor (single working electrode or dual working electrode), is configured to provide one or more data points, with which the continuous analyte sensor can be calibrated, without the use of a separate (e.g., external to the system) device and/or testing of reference fluids.

**[0683]** Accordingly, in preferred embodiments, the continuous analyte detection system includes a continuous analyte sensor (e.g., described elsewhere herein) and a reference analyte sensor. The continuous analyte sensor is configured to detect a first signal associated with a test analyte and a second signal is associated with a reference analyte. The reference sensor is configured to generate a reference signal associated with the reference analyte. The test analyte can be any analyte that the sensor is configured to continuously monitor in the host. For example, in some preferred embodiments, the test analyte is glucose. The reference analyte is an analyte other than the test analyte, which is substantially stable within the host. For example, in general, the concentration of the reference analyte (e.g., in the host's circulatory system) does not fluctuate rapidly. In some preferred embodiments, the reference analyte is oxygen ($O_2$), however, a variety of other analytes can be used. The reference analyte selected is an analyte that can be measured by both the continuous analyte sensor and the reference sensor.

**[0684]** The continuous analyte sensor can be any type of continuous analyte sensor, including a continuous analyte sensor having a single working electrode or dual-working electrodes. In some embodiments, the continuous analyte sensor is a single working electrode continuous analyte sensor configured to detect glucose, and the first signal is associated with glucose. In this embodiment, the working electrode is configured to detect the second signal (associated with the reference analyte). For example, in some embodiments, the sensor is configured to generate a signal associated with glucose when a +0.6mV potential is applied to the sensor. In some circumstances, the sensor can detect another analyte, if a different potential is applied thereto. For example, if a -0.6mV potential is applied, the sensor can detect $O_2$. Accordingly, in some embodiment, the system is configured to detect both glucose and $O_2$ (e.g., first and second signals) at the working electrode of the continuous analyte sensor, by switching the potential applied to the sensor. In some embodiments, the continuous analyte sensor includes an auxiliary electrode, which can be configured to detect the

reference analyte (second signal), as described herein with reference to "transport-measuring" electrodes.

**[0685]** In other embodiments, the continuous analyte sensor is a dual-working electrode continuous analyte sensor configured to detect glucose, and the first signal (detected by the working electrode disposed beneath an enzymatic portion of the membrane) is associated with glucose. In some embodiments, the system is configured to detect the second signal (associated with the reference analyte) using the dual-electrode sensor's first working electrode (**E1**, with enzyme) as described above. In other embodiments, the system is configured to detect the second signal (reference analyte) using the dual-electrode sensors second working electrode (**E2**, no enzyme), such as by applying a -0.6mV potential thereto. In this embodiment, the second signal associated with the reference analyte should not be confused with the signal detected by the second working electrode that is associated with non-analyte-related electroactive species that have an oxidation/reduction potential that substantially overlaps with the analyte's oxidation/reduction potential. For example, in some embodiments, the dual-electrode sensor is configured such that when a +0.6mV potential is applied to the second working electrode (**E2**, disposed beneath a non-enzymatic portion of the membrane) the signal generated is associated with the non-analyte-related electroactive species that have an oxidation/reduction potential that substantially overlaps with the analyte's oxidation/reduction potential; then, when a -0.6mV potential is applied to the second working electrode, the second working electrode detects the second signal (associated with the reference analyte).

**[0686]** In some embodiments, the continuous analyte sensor includes more than two working electrodes disposed beneath the membrane. In one exemplary embodiment, the sensor includes a first working electrode **E1** configured to generate a signal associated with the analyte (the first signal), a second working electrode **E2** configured to generate a second signal associated with the reference analyte, and a third working electrode **E3** configured to generate a signal associated with the non-analyte-related electroactive species that have an oxidation/reduction potential that substantially overlaps with the analyte's oxidation/reduction potential. In some embodiments, the sensor can include an additional working electrode (e.g., **E4, $E_n$)** configured to detect another reference analyte and/or to generate signals associated with non-analyte-related species that have oxidation/reduction potentials that overlap with that of another analyte, the reference analyte, another reference analyte, and the like.

**[0687]** In preferred embodiments, the reference sensor is not disposed beneath the analyte sensor's membrane and is configured to detect the reference analyte using any means known in the art, such as but not limited to electrochemical, enzymatic, chemical, physical, immunochemical, radiometric, and the like. In some preferred embodiments, the reference sensor is an optical sensing apparatus configured to detect the reference analyte. For example, the reference sensor can be an optical $O_2$ sensing apparatus configured to use one of a variety of optical detection methods known in the art and a described herein in the section entitled "Optical Detection."

**[0688]** In preferred embodiments, the system is configured such that the continuous analyte sensor and the reference sensor are disposed in the same local environment and are therefore simultaneously exposed to (e.g., contacted by/with) the sample (e.g., blood). For example, in some embodiments, the system can be configured such that the continuous analyte sensor is a wire sensor configured to extend into a catheter, and the reference sensor comprises an optical fiber that is configured both to detect the reference analyte and to extend into a catheter such that the detecting portion of the reference sensor is adjacent to the sensing portion of the continuous analyte sensor. In another exemplary embodiment, the continuous analyte sensor and the reference sensor are disposed within a connector; such as described in the section entitled "Multi-sensor apparatus." In still another exemplary embodiment, the continuous analyte sensor and the reference sensor are integrally formed on a vascular access device, such as on the *in vivo* portion of a catheter. For example, an optical fiber can be incorporated into the *in vivo* portion of the catheter during manufacture (e.g., such as *via* injection molding techniques known in the art) or by attaching the optical fiber with an adhesive, and subsequent the deposition of the continuous analyte sensor electrodes to the exterior surface of the *in vivo* portion of the catheter, as described herein.

**[0689]** In preferred embodiments, the system includes a processor module configured to process the second signal (associated with the reference analyte) and the reference signal to calibrate the first signal (associated with the analyte). For example, in some embodiments, the processor module uses the reference signal, which is generated by a sensor outside the membrane, to calibrate the second signal (generated under the membrane). Accordingly, shifts in baseline and/or sensitivity, which can arise over time during use of the sensor, are accounted for prior to calibration of the first signal (generated under the membrane). The processor configured to then calibrate the first signal (analyte signal) using the calibrated second signal, which generates a first signal substantially without a non-analyte signal component (and substantially unaffected by shifts in sensitivity (*m*) and/or baseline (*b*)).

**[0690]** As a non-limiting example, in one embodiment, the system includes a continuous glucose sensor configured to detect a first signal associated with glucose and a second signal associated with $O_2$. The system also includes an optical sensing apparatus configured to detect $O_2$. The system is configured such that the glucose sensor and the optical $O_2$ sensing apparatus can be exposed simultaneously to a sample. This method for calibration requires measurement of a second analyte that is already being monitored. For example, optical sensors are almost always used to measure $O_2$ in the host. In some embodiments, the glucose sensor's first working electrode generates both the analyte signal and the $O_2$ signal. In some embodiments, an electrode other than the first working electrode (e.g., a second or third

working electrode) is configured to generate the $O_2$ signal. Then the optical $O_2$ sensor can be used to calibrate the $O_2$ electrode (of glucose sensor system). Assuming there is a known relationship between the sensitivities of an electrode configured to generate a glucose signal and an electrode configured to generate an $O_2$ signal, then the sensor's glucose electrode can be calibrated by the $O_2$ electrode.

**[0691]** While not wishing to be bound by theory, it is believed that an analyte detection system configured to calibrate the analyte signal using a second/reference analyte provides a plurality of advantages. Primarily, calibration of a system of the preferred embodiments does not require input (manually or automatically) of reference data points from a secondary detection system (e.g., separate from the analyte detection system), such as a hand-held glucose meter. Similarly, no special IV bag, mechanical components or dedicated IV lines are required. A wide variety of analytes can be detected by both electrochemical means and a secondary means, such as optical detection methods. All of these advantages conflate to provide highly accurate, "plug-and-play" style continuous analyte detection system that is usable in a wide variety of settings.

Integrated sensor system

System Overview

**[0692]** As described above, tight control of glucose levels is critical to patient outcome in a critical care medical setting, especially for diabetic hosts. Maintaining tight glucose control with current technology poses an undue burden to medical personnel, due to time constraints and the extensive patient contact required. Reducing medical staff workload is a key component of improving patient care in this setting. The preferred embodiments disclose systems and methods to maintaining tight glucose control in the host while reducing and/or minimizing staff-patient interactions. Additionally, the preferred embodiments decrease testing intervals and improve sensor accuracy and reliability.

**[0693]** **Figs. 6** and **7** illustrate one preferred embodiment of the integrated sensor system **600** (e.g., for use at the bedside), which couples to the analyte sensor **14** (e.g., a glucose sensor) and vascular access device **12** (e.g., a catheter placed in a peripheral vein or artery) described above (see **Figs. 1A-1E),** and which includes at least one fluid reservoir **602** (e.g., a bag of calibration or IV hydration solution), a flow control device **604** (e.g., to control delivery of an infusion fluid **602a** from the reservoir to the host *via* the catheter), a local analyzer **608** and a remote analyzer **610.** In some embodiments, the analyte sensor is configured to reside within the catheter lumen **12a** (see **Figs. 1A-1E).** In some embodiments, the sensor is disposed within the catheter such the sensor does not protrude from the catheter orifice **12b.** In other embodiments, the sensor is disposed within the catheter such that at least a portion of the sensor protrudes from the catheter orifice. In still other embodiments, the sensor is configured to move between protruding and non-protruding dispositions. The analyte sensor and vascular access device used in the integrated sensor system **600** can be any types known in the art, such as but not limited to analyte sensors and vascular access devices described above, in the sections entitled "Applications/Uses" and "Exemplary Sensor Configurations." For convenience, the vascular access device **12** will be referred to as a catheter herein. However, one skilled in the art appreciates that other vascular access devices can be used in place of a catheter.

**[0694]** In some embodiments, at least one electronics module (not shown) is included in the local and/or remote analyzers **608, 610** respectively, for controlling execution of various system functions, such as but not limited to system initiation, sensor calibration, movement of the flow control device **604** from one position to another, collecting and/or analyzing data, and the like. In preferred embodiments, the components and functions of the electronics module can be divided into two or more parts, such as between the local analyzer and remote analyzer, as is discussed in greater detail in the sections entitled "Local Analyzer" and "Remote Analyzer."

**[0695]** In some embodiments, the flow control device **604** includes one or more valves and is configured to control fluid delivery to the host and sample take-up (e.g., drawing blood back into the catheter until at least the sensor's electroactive surfaces are contacted by the blood). In some embodiments, the sensor **14** dwells within the lumen **12a** of the catheter **12,** as described elsewhere herein. In some embodiments, wherein an internal calibration is performed, an infusion fluid (e.g., calibration solution **602a)** flows over the indwelling sensor **14** and is infused into the host. Generally, analyte in the solution **602a** can be measured when the sensor electroactive surfaces are in contact with the solution **602a.** In some embodiments, the measurements of the solution **602a** can be used to calibrate the sensor **14.** After calibration, the system is configured such that a sample (e.g., blood or other bodily fluid) contacts the sensor's electroactive surfaces (e.g., by drawing blood back into the catheter). When the sample contacts the electroactive surfaces, the sample's analyte concentration can be detected by the sensor **14.** When a sample is drawn back, the sample can then be returned to the host. In some embodiments, the integrated sensor system **600** cycles between calibration (e.g., measurement of a reference calibration solution) and measurement (e.g., of a sample, such as blood, glucose concentration). In some embodiments, the system **600** continues operation in this cyclical manner, until the system **600** is either disconnected from the host or turned off for a period of time (e.g., during movement of the host from one location to another). For example, in one embodiment, the system **600** cycles between the calibration and measurement steps from

about every 30 seconds or less to about every 2 hours or more. In another embodiment, the system **600** cycles between the calibration and measurement steps of from about every 2 minutes to about every 45 minutes. In still another embodiment, the system **600** cycles between the calibration and measurement steps from about every 1 minute to about every 10 minutes. In some embodiments, the user can adjust the time between steps. In some embodiments, the user can adjust the time between each step. In some embodiments, the system **600** can perform additional steps, such as but not limited to a flushing step, a keep vein open step (KVO), an extended infusion step, and the like. In some embodiments, the time is dependent upon sensors that detect a reference solution (e.g., calibration solution) and/or sample (e.g., blood) at the electroactive surfaces.

**[0696]** The integrated sensor system **600** of the preferred embodiments provides several advantages over prior art technology. Namely, in preferred embodiments, continuous analyte monitoring is enabled. When the analyte is glucose, continuous glucose monitoring enables tight glucose control, which can lead to reduced morbidity and mortality among diabetic hosts. Additionally, the medial staff is not unduly burdened by additional patient interaction requirements. Advantageously, there is no net sample (e.g., blood) loss for the host, which is a critical feature in some clinical settings. For example, in a neonatal intensive care unit, the host is extremely small and loss of even a few milliliters of blood can be life threatening. Furthermore, returning the body fluid sample to the host, instead of delivering to a waste container greatly reduces the accumulation of biohazardous waste that requires special disposal procedures. The integrated sensor system components, as well as their use in conjunction with an indwelling analyte sensor, are discussed in greater detail below.

Fluids

**[0697]** Referring to **Figs. 6** and **7**, in preferred embodiments, the integrated sensor system **600** includes at least one reservoir **602** that contains an infusion fluid **602a,** such as but not limited to reference (e.g., calibration), hydration and/or flushing solutions. For simplicity, the infusion fluid **602a** will be referred to herein as a solution **602a.** However, one skilled in the art recognizes that a wide variety of infusible fluids can be used in the embodiments discussed herein.

**[0698]** In some embodiments, the reservoir **602** includes a container such as but not limited to an IV bag. In other embodiments, the reservoir **602** can include two or more IV bags, or any other sterile infusion fluid container. In some embodiments, the reservoir **602** is a multi-compartment container, such as but not limited to a multi-compartment IV bag. If two or more solutions **602a** (e.g., calibration solutions, flush solutions, medication delivery solutions, etc.) are used, the solutions **602a** can be contained in two or more IV bags or in a multi-compartment IV bag, for example. In some embodiments, it is preferred to use a single solution **602a.** Use of a single solution **602a** for calibration, catheter flushing and the like simplifies the system **600** by reducing the complexity and/or number of system **600** components required for system **600** function. In some embodiments, two or more solutions **602a** are preferred, and can be provided by a multi-compartment IV bag or two or more separate reservoirs **602** (e.g., two or more bags, each containing a different solution **602a**). Advantageously, use of multiple solutions **602a** can increase system functionality 600 and can improve sensor accuracy.

**[0699]** Any infusion fluid (e.g., solution **602a**) known in the art can be used in conjunction with the present system **600.** In some embodiments, the solution **602a** is an analyte-containing solution that can be used as a reference or standard for sensor **14** calibration (generally referred to as a reference and/or calibration solution in the art). In some embodiments, a solution **602a** can be used as a flushing solution, to wash a sample off the sensor **14** and out of the catheter **12**. In some embodiments, two or more solutions **602a** (e.g., having different analyte concentrations) can used to provide two or more calibration measurements. In one exemplary embodiment, the analyte sensor **14** is a glucose sensor, and the solution **602a** contains dextrose or glucose at a concentration of from about 0 mg/dl to about 400 mg/dl. In preferred embodiments, the solution **602a** contains from about 75 mg/dl to about 200 mg/dl glucose. In more preferred embodiments, the solution **602a** contains from about 100 mg/dl to about 150 mg/dl glucose. In some embodiments, the solution **602a** is an isotonic saline solution. In some embodiments, the solution **602a** contains a sufficient concentration of an anticoagulant to substantially prevent blood clotting in and/or near the catheter **14.** In some embodiments, the solution **602a** contains a sufficient concentration of or antimicrobial to substantially prevent infection in and/or near the catheter. In one exemplary embodiment, the reservoir **602** is a 500 ml bag containing a sterile solution **602a** including 0.9% sodium chloride in water (e.g., normal saline), 2 IU/ml heparin and 100 mg/dl dextrose. In another exemplary embodiment, the reservoir **602** is a 500 ml bag containing heparinized saline.

**[0700]** In some embodiments, one, two or more solutions **602a** can be used in conjunction with the integrated sensor system **600.** For example, in some embodiments, two or more calibration solutions **602a** (e.g., solutions with different analyte concentrations) can be used. In one preferred embodiment, the analyte sensor **14** is a glucose sensor and the calibration solution **602a** includes a glucose concentration of from 0 mg/dl to about 300 mg/dl or more. In one exemplary embodiment, a single calibration solution **602a** (e.g., having a 100 mg/dl glucose concentration) can be used. In another exemplary embodiment, two calibration solutions **602a** (e.g., having 100 mg/dl and 0 mg/dl glucose concentrations) can be used. In other exemplary embodiments, three calibration (e.g., 0 mg/dl glucose, 75 mg/dl glucose and 300 mg/dl

glucose) solutions **602a** can be used. In still other embodiments, more than three calibration solutions **602a** can be used. In addition to calibration solutions **602a,** non-calibration solutions **602a** can be used in conjunction with the integrated sensor system **600,** such as but not limited to intravenously administered drugs, insulin, enzymes, nutritional fluids, and the like.

**[0701]** The solution **602a** can be provided to the user in a variety of ways, depending upon local hospital protocol and/or physician preference. In some embodiments, the solution **602a** is supplied pre-mixed (e.g., an IV bag containing sodium chloride, dextrose and heparin), such that fluid reservoir **602** can be connected to an infusion set and infused into the host with minimal effort. In other embodiments, one or more of the solution components **602a** can be provided separately, such that the final solution **602a** is prepared at the host's bedside, at the nurse's station or in the hospital pharmacy, for example. In one exemplary embodiment, the solution **602a** can be provided to the medical staff as a kit including a bag of sterile solution (e.g., water) and injectable sodium chloride, dextrose and heparin aliquots of sufficient quantity to prepare the final solution **602a.** The solution **602a** can be mixed at the bedside or at a location remote from the host, and then applied to the host and to the integrated sensor system **600.** In some embodiments, the reservoir **602** is a 500 ml or 1000 ml bag containing a sterile solution of heparinized saline and 100 mg/dl, 150 mg/dl or 200 mg/dl glucose.

**[0702]** In various preferred embodiments, the solutions **602a** are administered with standard IV administration lines, such as those commonly used today, such as a sterile, single-use IV set, referred to herein as tubing **606.** In some embodiments, the tubing **606** can be provided with the solution(s) **602a.** While in other embodiments, the tubing **606** can be provided separately from the solution(s) **602a** or other system components. Additional system **600** components that can be provided with the solution(s) **602a** include but are not limited to a sensor **14,** a catheter **12,** tubing **606,** a local analyzer **608,** wires/cables for hard-wire connections between system components, and the like.

**[0703]** In some embodiments, multiple solutions **602a** can be infused through a multi-lumen catheter **12,** such as but not limited to a two-lumen or three-lumen catheter. In some embodiments, the sensor **14** is disposed in one of the catheter's lumens **12a,** through which one or more calibration solutions **602a** can be passed, while other fluids (e.g., hydration fluids, drugs, nutritional fluids) to be delivered to the patient are infused through the other catheter **12** lumens **12a** (e.g., second, third or more lumens).

**[0704]** In some embodiments, the reservoir **602** is held by a support **612.** The support **612** can take many forms, such as an elevated support. In some embodiments, the support **612** is an IV pole, such those commonly used in medical care facilities. In some embodiments, the reservoir **602** is suspended on the support **612,** and the height of the reservoir **602** can be adjusted (e.g., raised or lowered) to modulate solution **602a** discharge from the reservoir **602.**

**[0705]** In some embodiments, the reservoir **602** and solution **602a** can be provided with one or more system **600** components, such as in a kit. In one exemplary embodiment, a kit including the components to mix the solution **602a** can include an analyte sensor **14** and a standard infusion set (e.g., catheter **12,** cannula, IV tubing **606,** etc.). In other embodiments, a kit can include a premixed solution **602a**, with an analyte sensor **14**. In various embodiments, a kit can contain instructions for use, such as for mixing the solution **602a** and applying it to the integrated sensor system **600.** Advantageously, providing either a pre-mixed solution **602a** or solution components with one or more system **600** components (e.g., sensor **14,** catheter **12,** tubing **606**, local analyzer **608**) can increase efficiency of medical care and provide ease of use to the nursing staff.

Flow Regulators

**[0706]** Still referring to **Figs. 6** and **7**, in some embodiments, a flow regulator **602b** controls the solution **602a** flow rate from the reservoir **602** to the flow control device **604**, which is described below. A variety of flow regulators can be used with the preferred embodiments, including but not limited to pinch valves, such as rotating pinch valves and linear pinch valves, cams and the like. In one exemplary embodiment, the flow regulator **602b** is a pinch valve, supplied with the IV set and located on the tubing **606** adjacent to and below the drip chamber. In some embodiments, a flow regulator **602b** controls the flow rate from the reservoir **602** to a flow control device **604**, which is described in the section entitled "Flow Control Device." In some embodiments, a flow regulator is optional; and a flow control device **604** controls the flow rate (e.g., from the reservoir **602** to the catheter **14**, described elsewhere herein).

Flow Control Device

**[0707]** In preferred embodiments, the integrated sensor system **600** includes a flow control device **604.** In some embodiments, the flow control device **604** is configured to regulate the exposure of the sensor **14** to the solution **602a** and to host sample (e.g., blood or other bodily fluid). In some embodiments, the flow control device **604** can include a variety of flow regulating devices, such as but not limited to valves, cams, pumps, and the like. In one exemplary embodiment, the flow control device **604** includes a simple linear pinch valve. In another exemplary embodiment, the flow control device **604** includes two or more linear pinch valves. In another exemplary embodiment, the flow control

device **604** includes one or more non-linear pinch valves. In another exemplary embodiment, the flow control device **604** includes a global valve. In still another exemplary embodiment, the flow control device **604** includes a gate valve, such as but not limited to a rising stem or non-rising-stem valve. In another exemplary embodiment, the flow control device **604** includes a butterfly valve or a ball valve. In still another exemplary embodiment, the flow control device **604** includes a pump, such as but not limited to volumetric infusion pumps, peristaltic pumps, piston pumps and syringe pumps. In still other exemplary embodiments, the flow control device **604** can be configured to vary the pressure at the reservoir **602,** such as but not limited to a pressure cuff around an IV bag and/or raising/lowering the reservoir adjust head pressure. In some embodiments, the flow control device **604** includes a gravity-fed valve. In still other embodiments, the flow control device **604** is configured to use flow dynamics at the catheter **12,** to regulate exposure to the sensor to solution or sample, as described elsewhere herein. Although some exemplary glucose sensors are described in detail herein, the system **600** can be configured to utilize a variety of analyte sensors including a variety of measurement technologies, such as enzymatic, chemical, physical, electrochemical, spectrophotometric, polarimetric, calorimetric, radiometric, and the like.

**[0708]** Referring now to a preferred embodiment wherein the sensor is an enzyme-based sensor, it is known to those skilled in the art that the rate of an enzymatic reaction is temperature dependent. Depending upon the enzyme, temperature reductions generally slow enzymatic reaction rates; temperature increases generally increase reaction rates. Since the analyte sensors **14** described in the preferred embodiment herein depend upon an enzyme (e.g., GOX) to detect the analyte (e.g., glucose) temperature changes during sensor calibration can result in artifacts on the sensor signal. For example, if the solution **602a** temperature is reduced (relative to body temperature), the enzymatic reaction will proceed at a reduced rate (relative to the rate at body temperature), causing the solution's analyte concentration to appear artificially low, which can result in improper sensor calibration. In some circumstances, changes in the relative temperatures of the area of the host's body surrounding the sensor (e.g., the blood contacting the sensor and the flesh surrounding the implanted sensor) and of the solution **602a** can be caused by the host moving the implant site, covering (or uncovering) an implant site with a blanket, application of a heating pad or ice to the implant site, and the like. In some circumstances, a high flow rate can cause large temperature fluctuations when the sensor is alternately exposed to blood and solution **602a.** For example, if the flow rate is sufficiently slow, the infusion fluid **602a** can be sufficiently warmed by the body before it contacts the sensor; thus the calibration measurements taken will be made at a temperature substantially similar to the temperature with the test measurements are taken in blood. If the flow rate is too fast, the infusion fluid **602a** will not warm up sufficiently, and the temperature will be too cold when the calibration measurements are taken, which can lead to improper sensor calibration. An improperly calibrated sensor can aberrantly measure the analyte concentration in the sample (e.g., blood from the host). Aberrant readings of sample analyte concentration can lead to improper treatment decisions by the medical staff and/or the host. The effects of temperature on enzymatic reaction rates can be mathematically described using a temperature coefficient. Signal artifacts caused by temperature-related reductions in enzyme reaction rate are referred to herein as temperature coefficient artifacts.

**[0709]** Generally, the host tissue in which the catheter **12** has been implanted surrounds an *in vivo* portion of the catheter **12**. In preferred embodiments, the flow control device **604** is configured to pass the solution **602a** through the catheter **12** at a rate such that the solution's temperature substantially equilibrates with the temperature of the surrounding host tissue. In one exemplary embodiment, the flow control device **604** maintains a flow rate of from about 0.5 μl/min or less to about 1.5 ml/min or more. In one preferred embodiment, the flow rate is from about 1 μl/min to about 1.0 ml/min. In one exemplary preferred embodiment, the flow rate is from about 0.01 ml/min to about 0.2 ml/min. In another exemplary preferred embodiment, the flow rate is from about 0.05 ml/min to about 0.1 ml/min. Advantageously, since the flow control device **604** infuses the solution **602a** at a rate sufficient to allow substantial temperature equilibration with the surrounding tissue, sensor **14** accuracy is improved and the integrated sensor system **600** has substantially no temperature coefficient artifacts.

**[0710]** In some alternative embodiments, a faster flow rate that does not allow for temperature equilibration is preferred. In such circumstances, measurement inaccuracies due to temperature coefficient can be generally eliminated mathematically using $b_{offset}$ and the calibration methods described in the section entitled "Systems and Methods for Processing Sensor Data."

**[0711]** In some embodiments, sample is taken up into the same catheter lumen **12a** through which the solution **602a** is infused into the host (described elsewhere herein). Thus, it is preferred that mixing of the sample and the solution **602a** is prevented. Similarly, it can be advantageous to detect when the sensor **14** is in contact with undiluted sample and/or undiluted solution. In some preferred embodiments of the integrated sensor system **600**, the flow control device **604** is configured to substantially prevent mixing of two or more fluids, such as but not limited to the solution **602a** and a host sample (e.g., blood). In preferred embodiments, mixing can be substantially prevented by a combination of factors, including specific gravity and flow rate. It is known that two solutions with different specific gravities tend not to mix, provided that the fluids are moved at a sufficiently slow rate (e.g., flow rate). Human whole blood has a specific gravity of about 1.05-1.06, while an infusion solution of 5% dextrose and 0.225% NaCl has a specific gravity of about 1.0189. Due to the difference in specific gravities, a blood sample and the solution **602a** tend to resist mixing within the tubing

**606** when the flow rate is sufficiently slow. In preferred embodiments, the sample and the solution **602a** are moved within the catheter lumen **12a** at a rate such that substantially no mixing occurs therebetween. In some embodiments, the flow rate is from about 0.001 ml/min or less to about 2.0 ml/min or more. In preferred embodiments, the flow rate is from about 0.01 ml/min to about 1.0 ml/min. In one exemplary preferred embodiment, the flow rate is from about 0.02 ml/min to about 0.35 ml/min. In another exemplary preferred embodiment, the flow rate is from about 0.0.02 ml/min to about 0.2 ml/min. In yet another exemplary preferred embodiment, the flow rate is from about 0.085 ml/min to about 0.2 ml/min.

**[0712]** In preferred embodiments, the flow control device **604** can include a variety of fluid flow-regulating devices known in the art. In some embodiments, the flow control device **604** includes one or more valves, such as but not limited to linear and non-linear roller valves, linear and non-linear pinch valves, bi-directional valves (either linear or non-linear), peristaltic rollers, cams, combinations thereof, and the like. In some other embodiments, the flow control device **604** is configured to generate sufficient "head pressure" to overcome the host's blood pressure such that the solution **602a** is infused into the host at a controlled rate; this can include elevating the fluid reservoir **602** (e.g., gravity fed) and using a valve to control the fluid flow rate out of the reservoir **602** and into the host. In one exemplary embodiment, the fluid flows at a maximum rate (e.g., about 6.25 ml/hr) such that a maximum fluid volume of about 150 ml/day can be infused into the host, however ranges much higher and/or lower can be implemented with the preferred embodiments.

**[0713]** In one exemplary embodiment, the flow control device **604** is a rotating pinch valve that has first and second positions. The valve can move between the two positions, for example, backward and forward, and thereby move fluids in and out of the catheter, as described in the section entitled "Flow Control Device Function." Namely, solution **602a** can be moved from the reservoir **602**, over the electroactive surfaces of the sensor **14** and into the host; and sample can be drawn up from the host, to cover the electroactive surfaces of the sensor **14,** and then pushed back into the host, by movement of the valve between the first and second positions.

**[0714]** In one exemplary embodiment, the flow control device includes a rotating pinch valve as described with reference to **Figs. 8A** through **8C.** Although **Figs. 8A** to **8C** describe one implementation of a rotating pinch valve that can be implemented with the sensor system, some alternatives include rotating pinch valves with multiple pinch surfaces, for example around the circumference of the rotateable axle **(Figs. 8A-8C, 804),** which enables the use of one valve for multiple infusion fluids (e.g., using multiple IV lines).

**[0715]** In some embodiments, the flow control device **604** includes one or more cams that regulate the flow rate. In one embodiment, the flow control device **604** includes a plurality of fixed orifices, which are opened and closed by the cams. As the cams are rotated, the flow increases and/or decreases in response. In one exemplary embodiment, the flow control device **604** includes three openings and three cams that mate with the openings (one cam per opening); fluid can flow through each opening at a given rate, *X* ml/min. Accordingly, when the cams close all three openings, flow is stopped. When one of the openings is opened, the fluid flows at *X* ml/min. If two openings are opened, fluid flows at 2*X* ml/min. Similarly, when the three openings are opened (e.g., by turning the cams such that they no longer close the openings), the fluid flows at **3***X* ml/min.

**[0716]** In another example, the flow control device **604** includes a plurality of cams and an equal plurality to tubes **606** passing through the cams, such that each cam can pinch closed the tube **606** that passes through it. In an exemplary embodiment, the cams are arranged such that they pinch and roll the tubing **606**, such that fluid is pushed into the host and sample taken up at pre-determined rates and times. For example, the flow control device **604** can include two cams, each having a tube **606** threaded therethrough. The cams are arranged such that each cam pinches and rolls the tubing **606** passing therethrough to push fluid into the host at one or more rates and to take up a blood sample.

**[0717]** In yet another example, the flow control device includes a rotating ball valve controlled by a motor, wherein the direction of the ball valve can be utilized to control a variety of functions, such as flow direction of the fluid.

**[0718]** As described in related herein, the sensor can be calibrated using more than one reference/calibration solution. For example, in some embodiments, the system is configured to calibrate using two reference solutions. In some embodiments, two solutions are used by metering the flow of one solution with the flow control device and intermittently stopping and intermittently manually injecting the second solution such that the solution contacts the sensor a sufficient period of time for calibration measurements to be taken. However, in preferred embodiments, the flow control device is configured to automate such "intermittent" calibrations (e.g., instead of manually injecting the calibration solution into the tubing) by actuating a secondary valve in a time-dependent manner, wherein the secondary valve in configured to meter the second calibration solution. The secondary valve can take on a variety of configurations, such as but not limited to a pinch valve, a ratchet valve, a cam, locking bearings or a ball valve (within the tubing).

**[0719]** As a non-limiting example, the secondary valve is a pinch valve that includes a ratcheting disk attached to the front of the flow control valve (e.g., *via* an axle), first and second arms, and includes a pin that can engage the tip of the second arm. In some preferred embodiments, the first and second arms are connected to each other by a joint, which includes a biasing means, such as a torsional spring, that pushes the second arm toward the ratcheting disk. The joint includes a stop pin, which limits the distance the second arm can move away from the ratcheting disk. The first arm includes a detent (e.g., a finger) that is pressed into the tubing (threaded through the flow control device) by another biasing means configured to push the first arm toward the ratcheting disk. Accordingly, in some embodiments, when the

flow control device rotates in the first direction the ratcheting disk is not engaged, and the first arm pinches the tube closed. In other embodiments, when the flow control device rotates in the first direction, the disk rotates with the flow control device such that the tip of the second arm rides up and over the pin of the secondary valve. When the flow control device rotates in the second direction, it is configured to engage the ratcheting disk and has two positions. As the flow control device rotates to the first position, blood is taken up into the catheter. At this point, the flow control device is configured to reverse and rotate in the first direction. However, a portion of the time, the flow control device rotates to the second position. As the flow control device rotates to the second position, the ratcheting disk remains engaged and the disk rotates far enough that the pin (on the disk) engages the tip of the second arm. Accordingly, the second arm is pushed away from the flow control device. The continued movement of the second arm is prevented by the stop pin. Since the second arm must continue to move away from the flow control device /ratcheting disk, the first arm is engaged by the second arm, and the entire structure is pushed away from the flow control device. Accordingly, the pinch on the tubing (e.g., by the detent) is relieved and the fluid in the tubing can flow. After a desired amount of fluid has flowed through the tubing, the pinch of the tubing (by the detent) is reapplied, by moving the flow control device/ratcheting disk in the first direction.

**[0720]** As another non-limiting example, in some embodiments the flow control device includes a cam configured to control the pinching/unpinching of the tubing. For example, a cam (e.g., weighted) is attached to the flow control device or another structure (e.g., a ratcheting disk) attached to the flow control device. The tubing is compressed (pinched) by a detent on an arm. The arm has a joint and is biased towards the flow control device via a biasing means. At desired time points, the flow control device is configured to move the cam to a second position (e.g., 180°), such that the arm is pushed upward (e.g., by the cam) and the pinch of the tubing is relieved. When the pinch of the tubing is relieved, the solution can flow through the tubing. The flow control device then returns the cam to its first position, so that the tubing is recompressed (pinched) and fluid flow is stopped.

**[0721]** In some embodiments, an electronics module (not shown) is incorporated into the flow control device **604,** to provide local control over flow control device function; in these embodiments, the flow control device function can be transmitted to the local and/or remote analyzer for processing. In other embodiments, a remote analyzer **610** and/or electronics module, such as but not limited to a computer system, controls the flow control device **604.** System **600** components that regulate the flow control device **604** are discussed in greater detail elsewhere herein.

**[0722]** In a further embodiment, the flow control device **604** is a computer controlled rolling pinch valve that acts on the exterior of sterile tubing **606** in order to control the gravity flow of a solution **602a** from an elevated fluid reservoir **602** into the host. In preferred embodiments, the flow control device **604** is configured to pinch and roll a small volume of tubing **606** such that a sample of host blood is drawn up into the catheter **12** (e.g., with a sensor **14** disposed therein) for analyte measurement, and to then push the sample back into the host with a solution (e.g., the calibration solution **602a).** In general, the flow control device **604** is configured to oscillate between drawing up a blood sample and allowing flow of the calibration solution **602a** at a predetermined rate. In some embodiments, the flow control device **604** includes at least one "hard stop" that ensures that the flow control device **604** does not move to a position that could endanger and/or injure the host, such as by draining the IV bag **602** of fluid **602a** or inappropriately (e.g., excessively) withdrawing blood, for example.

Tubing/Catheter

**[0723]** Referring again to **Figs. 6** and **7,** in preferred embodiments, the integrated sensor system **600** includes tubing **606** (e.g., sterile tubing configured for use in intravascular fluid infusion) and a catheter **12,** to deliver the solution **602a** from the reservoir **602** to the host. Generally, the tubing **606** and catheter **12** are sterile, single use devices generally used in medical fluid infusion, and may be referred to as an "infusion set." An infusion set may include additional components, such as but not limited to a cannula or needle for implanting the catheter, sterilization fluid (e.g., on a gauze pad) for cleaning/sterilizing the insertion site (e.g., the host's skin), tape, gauze, and the like. IV tubing is available in a variety of sizes and configurations, which find use in the preferred embodiments. For example, the tubing can be any size internal diameter, such as from about 0.5 mm to about 5 mm internal diameter. In various embodiments, the tubing can include a drip chamber and/or one or more access devices, such as but not limited to stopcocks, diaphragms and the like.

**[0724]** Catheters **12** are available in a variety of sizes and configurations. Catheters **12** for use in conjunction with an analyte sensor **14** are described in detail, elsewhere herein. Briefly, the catheter **12** can be any single- or multi-lumen catheter having a straight or divided tubing connector (e.g., straight-through, single shut off, double shut off, non-spill couplings, valves, T-connectors, Y-connectors, X-connectors, pinch clamps, leur locks, back-flow valves, and the like). In some embodiment, the catheter is configured for insertion into the venous side of the host's circulatory system. In other embodiments, the catheter is configured for insertion into the arterial side of the host's circulatory system, into either a peripheral or a central artery. In some embodiments, the catheter **12** is configured with an integrally formed sensor **14.** In alternative embodiments, a non-integral sensor **14** is configured for insertion into the catheter **12** after

catheter insertion. In some embodiments, the catheter **12** is a single lumen catheter that is configured for infusion of a fluid. In preferred embodiments, an indwelling sensor **14** is disposed within the catheter's lumen **12a.** In some embodiments, the catheter **12** and sensor **14** are provided to a user together. In other embodiments, the catheter **12** and sensor **14** are supplied separately. In an alternative embodiment, the catheter **12** is a multi-lumen catheter configured for infusion of two or more solutions. In preferred embodiments, a sensor **14** is disposed within one of the catheter's multiple lumens **12a.** For example, a calibration solution **602a** (e.g., 100 mg/dl glucose in saline) can be infused through the lumen **12a** in which the sensor **14** is disposed, while a hydration fluid (e.g., including a medication) can be infused through a second lumen. Advantageously, a dual lumen catheter **12** allows non-interrupted system use while other fluids are concurrently provided to the host.

**[0725]** In some embodiments, only the working electrode(s) of the sensor **14** are disposed within the catheter lumen **12a** and the reference electrode is disposed remotely from the working electrode(s). In other embodiments, the sensor **14** is configured to intermittently protrude from the catheter lumen **12a.**

Sample-Contacting Sensor

**[0726]** In preferred embodiments, the integrated sensor system **600** is configured such that at least the sensor's electroactive surfaces can be exposed to a sample and the sample's analyte concentration can be detected. Contacting the sensor **14** with the sample can be accomplished in a variety of ways, depending upon sensor/catheter configuration. A wide variety of catheter **12** and/or sensor **14** configurations can be implemented in the preferred embodiments, to expose the sensor's electroactive surfaces to a biological sample. In one exemplary embodiment, the catheter **12** is disposed in the host's peripheral vascular system, such as in a peripheral vein or artery, and a blood sample is taken up into the catheter **12** such that the blood contacts the sensor's electroactive surfaces. In another exemplary embodiment, the catheter **12** can be disposed in the host's central vascular system or in an extracorporeal blood flow device, such as but not limited to an arterial-venous shunt, an extravascular blood-testing apparatus, a dialysis machine and the like, wherein blood samples can be taken up into the catheter **12** such that at least the sensor's electroactive surfaces are contacted by the drawn up blood sample.

**[0727]** In one exemplary embodiment, the sensor **14** is configured to reside within the catheter lumen **12a** (e.g., not protrude from the catheter tip); and the integrated sensor system **600** is configured to draw back a sample into the catheter lumen **12a** such that at least the sensor's electroactive surfaces are contacted by the sample. In some embodiments, the sensor **14** is a small-structured sensor having a width of less than about 1 mm. In one preferred embodiment, the sensor has a width of less than about 0.4 mm. In a more preferred embodiment, the sensor has a width of less than about 0.2 mm. In some embodiments, the catheter **12** has an internal diameter of from about 0.2 mm or less to about 2.0 mm or more, preferably from about 0.5 mm to about 1.0 mm. In some embodiments, the sensor **14** is configured such that its electroactive surfaces are at or adjacent to its tip, and the flow control device **604** is configured to take up sample into the catheter lumen **12a** until the sample covers at least the electroactive surfaces. In some embodiments, the electroactive surfaces are distal from the sensor's tip and sample is drawn farther back into the catheter lumen **12a** until the sample covers the electroactive surfaces. In some embodiments, the tip of the sensor is disposed about 3 cm, 2 cm, or 1 cm or less from a tip of the catheter.

**[0728]** In some embodiments, the sample taken up into the catheter's lumen **12a** covers only a portion of the sensor's *in vivo* portion. In other embodiments, the sample taken up into the catheter's lumen **12a** covers the entire *in vivo* portion of the sensor **14.** In some embodiments, a sample volume of from about 1 μl or less to about 2 ml or more is taken up into the catheter **12** and is sufficient to cover at least the electroactive surfaces of the sensor **14.** In some preferred embodiments, the sample volume is from about 10 μl to about 1 ml. In some preferred embodiments, the sample volume is from about 20 μl to about 500 μl. In other preferred embodiments, the sample volume is from about 25 μl to about 150 μl. In more preferred embodiments, the sample volume is from about 2 μl to about 15 μl.

**[0729]** In preferred embodiments, the sample taken up into the catheter's lumen **12a** remains within the *in vivo* portion of the catheter **12.** For example, in some embodiments, the sample is not drawn so far back into the catheter **12** that it enters the *ex vivo* portion of the catheter **12,** the tubing **606** or the reservoir **602.** In some embodiments, however, the sample can be drawn back as far as the catheter but not into the IV tubing. In some embodiments wherein the catheter **12** is implanted in a host, the blood sample never leaves the host's body (e.g., a plane defined by the host's skin). In some embodiments wherein the catheter **12** is implanted in an extracorporeal device, the sample does not substantially exit the extracorporeal device. In preferred embodiments, wherein blood is taken up into the catheter **12,** the blood is returned to the host (or extracorporeal device), which is described elsewhere herein. In preferred embodiments, the sample is blood taken up from the host's circulatory system and into the catheter **12** disposed within the circulatory system.

**[0730]** In another exemplary embodiment of the integrated sensor system, the sensor is configured to protrude from the catheter's orifice **12b,** at least intermittently. In preferred embodiments, the sensor is configured to protrude sufficiently far out of the catheter's lumen **12a** (e.g., into the circulatory system proper) that the sensor's electroactive surfaces are contacted by sample (e.g., blood). In a further embodiment, the sensor is configured to intermittently protrude from the

catheter orifice **12b,** such as by moving back and forth, such that the electroactive surfaces are alternately disposed within the catheter **12** and outside of the catheter **12.** In one exemplary embodiment of a catheter is implanted in a host's vein, calibration solution **602a** is provided within the catheter **12** such that the sensor **14** is disposed within the catheter **12,** the sensor **14** is contacted by the calibration solution **602a** and calibration measurements can be obtained periodically, when the sensor **14** (e.g., electroactive surfaces) is moved outside of the catheter **12,** the sensor **14** is contacted by blood and blood analyte measurements can be obtained.

**[0731]** In some embodiments of the integrated sensor system **600,** the catheter **12** and sensor **14** are configured to take advantage of flow dynamics within the host's vascular system. By taking advantage of flow dynamics, the system can be simplified, such that the flow control device functions mainly to allow or block the flow of calibration solution.

**[0732]** **Fig. 9** is a cut-away illustration of one exemplary embodiment, in which a catheter **12** is implanted in a host's vessel **906,** such as but not limited to an artery or vein. The catheter **12** includes a sidewall **904** that can be configured to include one or more holes **902** (e.g., orifices or openings configured for fluid passage, such as from the exterior sidewall surface into the catheter lumen **12a).** The catheter **12** can be inserted into the host's vein (or artery, or an extracorporeal circulatory device) such that the catheter points either in the direction of blood flow (antegrade) or against the direction of blood flow (retrograde). The catheter is configured such that in an antegrade position, blood flows into the catheter lumen **12a** *via* the holes **902** and then out of the catheter orifice **12b.** In a retrograde position, blood enters the catheter lumen **12a** *via* the catheter orifice **12b** and flows out of the lumen through the holes **902.** In some embodiments, the sensor **14** can be disposed within the catheter lumen **12a** such that blood flowing between the holes **902** and the orifice **12b** contacts at least the sensor's electroactive surfaces. In some embodiments, the sensor **14** is configured to be substantially immobile within the lumen **12a,** while in other embodiments the sensor **14** is configured to be substantially moveable within the lumen **12a,** as described in more detail elsewhere herein.

**[0733]** Generally, the holes **902** can be placed in any location on the catheter's sidewall **904.** In some embodiments, the holes **902** can be located near or adjacent to the catheter orifice **12a.** In other embodiments, the holes **902** can be placed remotely from the catheter orifice **12a.** The size, shape and number of holes **902** can be selected to optimize the sample volume and flow rate through the catheter lumen **12a.** For example, in some embodiments, the holes **902** are round, ellipsoid, rectangular, triangular, star-shaped, X-shaped, slits, combinations thereof, variations there of, and the like. Similarly, in some embodiments, the catheter **12** can have from 1 to about 50 or more holes **902.** In other embodiments, the catheter can have from 2 to about 10 or more holes **902.**

**[0734]** In some alternative embodiments, the catheter includes at least one size wall orifice in place of an end tip orifice, which allows selective exposure of the sensor to the host's biological sample there through. A variety of alternative catheter configurations are contemplated in conjunction with the preferred embodiments.

**[0735]** In one exemplary embodiment of the integrated sensor system **600,** the flow control device **604** is configured to intermittently block the infusion of solution **602a** through the catheter **12,** which is configured with side holes **902** as described above. Additionally, the analyte sensor is disposed within the catheter lumen **12a** such that sample passing between the side holes **902** and the catheter orifice **12b** bathes the sensor's electroactive surfaces, during which time an analyte measurement can be obtained. When the flow control device **604** does not block infusion, the solution **602a** contacts the sensor's electroactive surfaces; and calibration measurements can be taken.

**[0736]** In some embodiments, a solution **602a** can be infused into the catheter **12** at a rate such that the flow of sample between the holes **902** and the orifice **12b** is substantially blocked and at least the electroactive surfaces are bathed in the solution **602a** (e.g., undiluted solution). In preferred embodiments, the sensor **14** can be calibrated while it is bathed in the undiluted solution **602a.**

**[0737]** In preferred embodiments, the sensor **14** is a small-structured sensor with at least one electrode, such as a working electrode, as described elsewhere herein. In some embodiments, the sensor **14** has two or more electrodes, such as but not limited to working, reference and counter electrodes. In some embodiments, the sensor **14** includes a reference electrode disposed remotely from the working electrode, as discussed elsewhere herein. In some embodiments, the sensor **14** includes two or more electrodes that are separated by an insulator, such as described in U.S. Patent Publication No. US-2007-0027385-A1.

**[0738]** In preferred embodiments, the electrode is a fine wire, such as but not limited to a wire formed from platinum, iridium, platinum-iridium, palladium, gold, silver, silver chloride, carbon, graphite, gold, conductive polymers, alloys and the like. In some exemplary embodiments, the sensor **14** includes one or more electrodes formed from a fine wire with a diameter of from about 0.001 or less to about 0.010 inches or more. Although the electrodes can by formed by a variety of manufacturing techniques (bulk metal processing, deposition of metal onto a substrate, and the like), it can be advantageous to form the electrodes from plated wire (*e.g.*, platinum on steel wire) or bulk metal (*e.g.*, platinum wire). It is believed that electrodes formed from bulk metal wire provide superior performance (*e.g.*, in contrast to deposited electrodes), including increased stability of assay, simplified manufacturability, resistance to contamination (*e.g.*, which can be introduced in deposition processes), and improved surface reaction (*e.g.*, due to purity of material) without peeling or delamination.

**[0739]** In some embodiments, one or more electrodes are disposed on a support, such as but not limited to a planar

support of glass, polyimide, polyester and the like. In some exemplary embodiments, the electrodes include conductive inks and/or pastes including gold, platinum, palladium, chromium, copper, aluminum, pyrolitic carbon, composite material (e.g., metal-polymer blend), nickel, zinc, titanium, or an alloy, such as cobalt-nickel-chromium, or titanium-aluminum-vanadium, and are applied to the support using known techniques, such as but not limited to screen-printing and plating. Additional description can be found in U.S. Patent No. 7,153,265, U.S. Patent Publication No. US-2006-0293576-Al, U.S. Patent Publication No. US-2006-0253085-A1, U.S. Patent No. 7,003,340, and U.S. Patent No. 6,261,440.

**[0740]** In some embodiments, an optional redundant sensor can be disposed within the catheter lumen, in addition to the sensor **14** described elsewhere herein. In one exemplary embodiment, a sensor **14** and a redundant sensor are disposed within the lumen of a sensor implanted in a host's peripheral vein, such that the electroactive surfaces of the sensor **14** are more proximal to the catheter orifice **12b** than the electroactive surfaces of the redundant sensor; wherein blood is taken up into the lumen **12a** such that the electroactive surfaces of both the sensor **14** and the redundant sensor are contact by the blood; such that analyte can be detected by both the sensor **14** and the redundant sensor and the redundant sensor measurements are used by the system **600** to confirm the sensor's **14** measurements. In a further embodiment, both the sensor **14** and the redundant sensor are intermittently concurrently contacted by the solution **602a** such that both the sensor **14** and the redundant sensor can take calibration measurements of the solution **602a,** wherein the calibration measurements of the redundant sensor are at least used to confirm the calibration measurements of the sensor **14.** In another embodiment, the calibration measurements from both the sensor **14** and the redundant sensor are used to calibrate the sensor **14.**

Local Analyzer

**[0741]** Referring to **Figs. 6** and **7,** in some embodiments, the integrated sensor system **600** includes a local analyzer **608** configured to operably connect to a remote analyzer **610.** In some embodiments, the local analyzer **608** is proximal to an analyte sensor **14** and the remote analyzer **610** is configured to operably connect to the local analyzer. However, alternative configurations are possible, such as the analyte sensor **14** can be operably connected to both the local and remote analyzers **608, 610** respectively. The remote analyzer **610** of the preferred embodiments is discussed below. In various embodiments, one or more functions of the local analyzer **608** can be transferred to the remote analyzer, as is appreciated by one skilled in the art. Likewise, in some embodiments, one or more functions of the remote analyzer **610** can be incorporated into the local analyzer **608.** In further embodiments, functions of the local and/or remote analyzers **608, 610** can be disposed in one, two, three or more physical bodies (e.g., separate housings), depending upon the integrated sensor system **600** configuration and/or component combinations. For example, in one embodiment, the local analyzer **608** includes a potentiostat, a power source (e.g., battery or connection to an electrical source), and data storage; and the local analyzer **608** is configured such that the potentiostat is disposed on the sensor's fluid coupler **20** and the remaining local analyzer **608** components are disposed elsewhere between the local analyzer **608** and the remote analyzer **610** (e.g., connected by wiring).

**[0742]** Operable connections between the local and remote analyzers **608, 610** and the analyte sensor **14** can be accomplished by a hard wire (e.g., USB, serial), RF communication, IR communication, and the like. In some embodiments, operable connections include a connector known in the art, such as but not limited to mating plug and socket units, screw connectors, clips and the like. In some embodiments, the connectors are separable. In other embodiments, the connectors are inseparable. In some embodiments, the connectors include a lock, to prevent inadvertent disconnection. In some embodiments, the local analyzer can be isolated from the remote analyzer by an isolation transformer.

**[0743]** In some embodiments, the local analyzer **608** is operably connected to the sensor **14** (e.g., the sensor electrode(s)), such as by a wire connection. A detailed description of electronic components and configurations is described elsewhere herein, for example, in the section entitled "Sensor Electronics." In some embodiments, the local analyzer **608** is disposed on or adjacent to the sensor, such as on the sensor fluid coupler **20.** In one exemplary embodiment, the sensor's fluid coupler **20** includes a local analyzer housing that includes at least a potentiostat. In some embodiments, the housing can include a battery and electronics, such that the sensor **14** can be powered, and data can be collected and/or transmitted to additional system electronics (e.g., electronics units disposed remotely from the sensor, such as on the host's arm, on the host's bed and in the remote analyzer, and the like). In some embodiments, the local analyzer **608** includes a small housing that is connected to the sensor **14** *via* a short wire (e.g., from about 1 cm or less to about 10 cm or more) and is taped to the host's skin, such as adjacent to the catheter's insertion site on the host's arm or hand. In a further embodiment, the local analyzer **608** includes a connector, such as but not limited to a "plug" configured to mate with a "socket" wired to the sensor **14,** such that an electrical connection can be made between the local analyzer **608** and the sensor **14.** In another embodiment, the sensor **14** includes a cable having a plug configured to connection to the local analyzer **608** *via* a socket. In still another embodiment, both the sensor **14** and the local analyzer **608** include cables configured to mate with each other *via* a plug and socket mechanism. Advantageously, a detachable configuration allows catheter/sensor insertion without a cumbersome connection to the local analyzer **608** as well as re-use of the local analyzer **608.** In an alternative exemplary embodiment, the local analyzer **608** is permanently connected to the

sensor **14** and cannot be disconnected therefrom; a single use, permanently connected configuration can simplify application to the host, can reduce the possibility of cross-contamination between hosts, does not require cleaning and/or sterilization between hosts, and can reduce operator error during application to the host.

**[0744]** In preferred embodiments, the local analyzer **608** includes at least the minimal electronic components and/or programming required to energize the sensor **14** and collect data therefrom, such as but not limited to a potentiostat. However, in some embodiments, the local analyzer **608** includes additional electronic components that can be programmed to analyze one or more components of the collected raw signal, or to store data, calibration information, a patient ID and the like. In one exemplary embodiment, the local analyzer **608** includes a potentiostat and a battery back up. The battery back up can maintain a potential on the sensor and store data (calibration and/or collected host data) for brief periods of time when the electronics can be disconnected, such as when the host is moved from one location to another. In one exemplary embodiment, the local analyzer **608** is disposed on or adjacent to the sensor **14** and is configured such that the host can be connected to a first remote analyzer **610** at one station, and then disconnected from the first remote analyzer **610,** moved to a new location and connected to a second remote analyzer **610** at the new location, and the local analyzer **608** retains sufficient data that the system **600** functions substantially without initialization or substantial delay upon connection to the new (second) remote analyzer **610.** In another example, the host can be disconnected from the first remote analyzer **610,** taken to another location for a procedure (e.g., for surgery, imaging, and the like) and then reconnected to the first remote analyzer **610** upon return to the original location without substantial loss of system **600** function upon reconnection.

**[0745]** In some embodiments, the local analyzer **608** includes two or more parts, such that only the potentiostat is disposed on or adjacent to the sensor **14** (e.g., sensor fluid coupler **20**) or the catheter (e.g., catheter connector **18);** other portions of the local analyzer **608** can be disposed remotely from the host, such as in a separate housing wired to the sensor and to the remote analyzer. In one exemplary embodiment, the two parts of the local analyzer **608** can be separated (e.g., unplugged) such that the host can be moved and the local analyzer **608** portion that is attached to the host goes with the host while the remaining portion stays with the remote analyzer **610.**

**[0746]** In still other embodiments, all sensor electronics components are disposed remotely from the host, such as in the remote analyzer **610.** For example, the sensor **14** can include an appropriate connector, plug and/or wiring to connect the sensor **14** to the remote analyzer **610,** which powers the sensor **14,** collects raw data from the sensor **14,** calibrates the sensor **14,** analyzes and presents the data, and the like. In one example, the sensor **14** includes a cable of sufficient length to permit plugging the sensor **14** into a remote analyzer **610** disposed at the host's bedside.

**[0747]** In still other embodiments, the local analyzer **608** can be incorporated into the remote analyzer **610,** such as housed in the same body as the remote analyzer **610,** for example. In one exemplary embodiment, both the local and remote analyzers **608, 610** are disposed in a housing attached to a support **612** (e.g., connected to an IV pole, placed on a bedside table, connected to the wall, clamped to the head of the host's bed) and connected to the analyte sensor *via* a wire or cable. In some embodiments, the cables/wires (e.g., for connecting the sensor to the local analyzer and/or the remote analyzer, and/or connecting the local analyzer to the remote analyzer) can be provided in the IV tubing set.

Remote analyzer

**[0748]** As discussed in the section entitled "Local Analyzer," the integrated sensor system **600** includes a remote analyzer **610. In** preferred embodiment, the remote analyzer **610** is configured to at least communicate with the local analyzer **608** and can be configured to control the flow control device **604** described in the sections entitled "Flow Control Device," and "Flow Control Device Function." Generally, the remote analyzer **610** is powered from a standard 120 VAC wall circuit or other suitable power source, for example. In some embodiments, the remote analyzer **610** is disposed at the host's bedside and can be configured to be disposed on a support **612,** such as but not limited to, mounted a mobile IV drip pole, attached to the wall, clamped to the host's bed, or sitting on a table or other nearby structure.

**[0749]** In preferred embodiments, the remote analyzer **610** includes a display, such as but not limited to a printout, an LED display, a monitor, a touch-screen monitor and the like. In some embodiments, the remote analyzer **610** includes both a hard copy display, such as a printer configured to print collected data, and a monitor. In some embodiments, the remote analyzer **610** is a programmable touch-screen panel PC configured to have different "screens" and "buttons" for control of system components (e.g., the sensor **14,** the flow control device **604,** etc.) and to display data, such as but not limited to host identification and condition, host food intake, medication schedules and dosage information, sensor identification, raw data, processed data, calibration information, and the like, such as in tables and/or graphs. In further preferred embodiments, the remote analyzer **610** is configured to be programmed, such that the operator can initiate system functions such as IV fluid line priming, starting and/or stopping the flow control device **604,** select among two or more solutions (e.g., between glucose concentrations), select the mode of data delivery (e.g., printer or on-screen), send data to a central location (e.g., the nurse's station or medical records), set alarms (e.g., for low and high glucose), and the like.

**[0750]** In some embodiments, the system **600** is configured to integrate with (e.g., be used in conjunction with) third

party medical devices, such as but not limited to a pulse-oxygen meter, a blood pressure meter, a blood chemistry machine, and the like. In such embodiments, the local and/or remote analyzers **608, 610** can be configured to communicate with the third party medical devices, such as but not limited to a patient monitor.

Flow Control Device Function

**[0751]** In some embodiments, the remote analyzer **610** controls the function of the flow control device **604.** In some embodiments, the flow control device includes electronics configured to control the flow control device. The flow control device **604** can be configured to perform a number of steps of operation, which are discussed below. Depending upon the system configuration and physician preferences, in some embodiments, one or more of the steps can be performed. In some embodiments, all of the steps are performed. In some embodiments, the steps of operation can be performed in the order in which they are presented herein. In other embodiments, the order of steps of operation can be varied (e.g., repeated, omitted, rearranged), depending upon various parameters, such as but not limited to the calibration solution **602a** selected, the particular infusion set selected, catheter **12** size, host condition, analyte of interest, type of sample and location of sample collection, integration with third party devices, additional infusion of fluids and the like.

**[0752]** **Figs. 8A** through **8C** are schematic illustrations of a flow control device in one exemplary embodiment, including its relative movement/positions and the consequential effect on the flow of fluids through the sensor/catheter inserted in a host. In general, steps performed by the flow control device **604,** include the steps of: contacting the sensor **14** with calibration solution **602a** (including sensor calibration) and contacting the sensor with a biological sample to be measured. In some embodiments, additional steps can be taken, such as but not limited to keep a vein open (KVO) step and a wash step. In the exemplary embodiment presented in **Figs 8A** though **8C**, the flow control device **604** is a roller valve configured to move between at least two positions, **810** and **812,** respectively. Movement of the flow control device **604** between positions **810** and **812** effectively concurrently moves the pinch point **808** (e.g., the point at which tubing **606** is pinched) between positions **810** and **812.** Additional flow control device positions are discussed below.

**[0753]** The top of **Figs. 8A** through **8C** are schematic drawings illustrating positions of the flow control device **604.** The bottom of **Figs. 8A** through **8C,** are a cut-away views of an implanted catheter **12,** including an indwelling sensor **14,** illustrating the corresponding activity at the implantation site, in response to movements of the flow control device **604.** For simplicity, for purposes of discussion only, it is assumed that the catheter **12** is implanted in a host's vein, that the sensor **12** does not protrude from the catheter's orifice **12b** and that the catheter **14** does not include side holes **902.** However, one skilled in the art appreciates that the catheter **14** could be implanted into any vessel of the host or into a variety of extracorporeal devices discussed elsewhere herein.

Step One: Contacting Sensor with Calibration Solution

**[0754]** In general, the system is configured to allow a calibration solution to contact the sensor using a flow control device such as a pump, valve or the like. In some embodiments, such as shown in **Figs. 8A** through **8C,** the flow control device **604** is a valve configured with a first structure **802** and a second structure **806.** For convenience, the first structure **802** is depicted as a roller connected to a rotateable axle **804,** however any flow control device such as described in the section entitled "Flow Control Device," can be configured to utilize the concepts and/or functions described herein. In general, when the flow control device is a valve, the valve is configured to allow no flow, free flow and/or metered flow through movement of the valve between one or more discreet positions.

**[0755]** In the embodiment shown in **Figs. 8A** through **8C,** the flow control device **604** is configured such that a tube **606** threaded between the first and second structures **802**, **806** (e.g., between the roller and the surface against which the roller presses) is compressed substantially closed. For convenience, the compressed location on the tubing is referred to herein as the "pinch point" **808.** In some embodiments, the flow control device **604** is configured such that the pinch point is moved along the tubing, either closer to or farther from the host. As the pinch point **808** is moved closer to the host, the tube **606** is progressively compressed, causing fluid (e.g., solution **602**) to be pushed into the host's vascular system (see the corresponding illustration of the sensor within the host's vessel at the bottom of **Fig. 8A**), at the catheter **12** implantation site. Conversely, as the pinch point **808** is moved away from the host, the portion of tubing **606** on the host side of the pinch point **808** progressively expands, causing sample (e.g., blood) to be drawn up into the catheter lumen **12a.** In an alternative embodiment, the flow control device **604** is configured such that the pinch point is substantially stationary and the first and second structures selectively compress the tubing at the pinch point (e.g., the tube **606** is either pinched fully closed or is fully open), which either stops or allows the flow of solution **602a.**

**[0756]** In the exemplary embodiment shown in **Fig. 8A** (bottom), the catheter **12** is implanted in the host's vein **906** (or artery), as described elsewhere herein. A sensor **14** is disposed with the catheter **12.** The catheter **12** is fluidly connected to a first end of tubing **606** that delivers the solution **602a** to the catheter **12.** The solution **602a** can move out of the catheter **12** and a sample of blood **814** can move in and out of the catheter **12,** *via* the catheter's orifice **12b.** In some alternative embodiments, the catheter **12** includes optional sidewall holes **902** (see Fig. 9, described elsewhere

herein) and the solution **602a** and blood can move in and out of the catheter **12** *via* the sidewall holes **902** and the catheter orifice **12b.** In some alternative embodiments, the sensor is configured to move in and out of the catheter. In some embodiments, the catheter orifice **12b** is disposed in the sidewall **904** (e.g., near the catheter's tip) instead of at the tip. Tubing 606 is fluidly connected to the reservoir **602** on a second end (see **Figs. 6** and **7**).

**[0757]** Referring now to a calibration phase to be performed by the exemplary valve of **Fig. 8A**, in preferred embodiments, the flow control device **604** is configured to perform a step of contacting the sensor **14** with solution **602a,** wherein the flow control device **604** moves from position **810** to position **812** (e.g., forward, toward the host/catheter). When the flow control device **604** moves from position **810** to position **812,** the pinch point **808** is moved from position **810** to position **812.** As the pinch point **808** is moved from position **810** to position **812,** a first volume of the calibration solution **602a** is pushed through the tubing **606,** toward the catheter **12.**

**[0758]** Referring again to the bottom of **Fig. 8A**, a second volume of the solution **602a,** which is substantially equal to the first volume, is pushed into the host's vein **906,** in response to the first volume of solution **602a** moving toward the host. As the second volume of solution **602a** is pushed through the catheter **12** and into the host's vein the second volume contacts (e.g., bathes) the analyte sensor **14,** including the analyte sensor's electroactive surfaces. In some embodiments, the volume (e.g., the first and second volumes of fluid) moved is from about 3 μl or less to about 1 ml or more. In some preferred embodiments, the volume is from about 10 μl to about 500 μl, or more preferably from about 15 μl to about 50 μl. In general, the volume of fluid pushed through the catheter in a particular phase (e.g., calibration phase) is dependent upon the timing of the phase. For example, if a long phase, such as a 20 minute calibration phase (e.g., as compared to a shorter 5 minute phase) were selected, the volume of fluid pushed during the long phase would be 4X greater than the volume of fluid pushed during the shorter phase. Accordingly, one skilled in the art appreciates that the above described ranges of fluids infusion can be increased and/or decreased simply be increasing or decreasing the measurement phase and/or intervals (i.e., timing). In preferred embodiments, the fluid is moved at a flow rate that is sufficiently slow that the calibration solution's temperature substantially equilibrates with the temperature of the tissue surrounding the *in vivo* portion of the catheter and/or temperature of bodily fluid (e.g., blood), such that the temperature of the calibration solution and the temperature of the blood are substantially the same. In preferred embodiments, the flow rate is from about 0.25 μl/min or less to about 10.0 ml/min or more. In one exemplary embodiment, the flow control device **604** maintains a flow rate from about 0.5 μl/min or less to about 1.5 ml/min or more. In one preferred exemplary embodiment, the flow rate is from about 1 μl/min to about 1.0 ml/min. In one exemplary preferred embodiment, the flow rate is from about 0.01 ml/min to about 0.2 ml/min. In another exemplary preferred embodiment, the flow rate is from about 0.05 ml/min to about 0.1 ml/min.

**[0759]** In some embodiments, the system is configured such that the speed of the movement between the first and second discreet positions is regulated or metered to control the flow rate of the fluid through the catheter. In some embodiments, the system is configured such that the time of movement between the first and second discreet positions is from about 0.25 to 30 seconds, preferably from about 0.5 to 10 seconds. In some embodiments, the system is configured such that an amount of pinch of the tubing regulates the flow rate of the fluid through the catheter. In some embodiments, the fluid flow is regulated through a combination of metering and/or pinching techniques, for example. Depending on the type of flow control device (e.g., valve), a variety of methods of metering and/or regulating the flow rate can be implemented as is appreciated by one skilled in the art.

**[0760]** Preferably, the sensor is configured to measure a signal associated with the solution (e.g., analyte concentration) during the movement of the flow control device from position **810** to position **812** and/or during contact of the sensor **14** with the solution **602a.** Electronics, such as an electronic module included in either the local or remote analyzer **608, 610** controls signal measurement and processing, such as described in more detail elsewhere herein.

**[0761]** In general, a calibration measurement can be taken at any time during the flow control device **604** movement from position **810** to position **812,** and including a stationary (stagnant) time there after. In some embodiments, one or more calibration measurements are taken at the beginning of the flow control device **604** movement from position **810** to position **812.** In other embodiments, one or more calibration measurements are taken at some time in the middle of the flow control device **604** movement from position **810** to position **812.** In some embodiments, one or more calibration measurements are taken near the completion of the flow control device **604** movement from position **810** to position **812.** In some embodiments, one or more calibration measurements are taken after completion of the flow control device **604** movement from position **810** to position **812.** In still other embodiments, the flow control device is positioned such that fluid can flow followed by positioning the flow control device such that there is no fluid flow (e.g., 0 ml/min) during the calibration measurement. In preferred embodiments, one or more calibration measurements are taken when the temperature of the solution **602a** has substantially equilibrated with the temperature of the tissue surrounding the *in vivo* portion of the implanted catheter **12.** Processing of calibration measurements and sensor calibration are described elsewhere herein.

**[0762]** As a non-limiting example, in some embodiments, the sensor can be calibrated using one or more reference solutions. For example, if the analyte is glucose, a suitable reference (e.g., calibration solution) is saline containing 0, 25, 50, 75, 100, 125, 150, 175, 200, 225, 250, 275, 300 mg/dl glucose or more. In some embodiments, two or more

such reference solutions can be used to calibrate the sensor. In some embodiments, a baseline value of the sensor can be obtained by generating a signal when the sensor is exposed to a 0 mg/dl reference solution (e.g., 0 mg/dl analyte). In some embodiments, updated baseline values are continuously obtained by repeatedly exposing the sensor to the 0 mg/dl reference solution, such as every 1, 2, 3, 4, 5, 10, 20, 30, 40, 60 or more minutes. In some embodiments, updated baseline values are continuously obtained by exposing the sensor to the 0 mg/dl reference solution for periods of time and continuously collecting baseline values. For example, the sensor can be exposed to the 0 mg/dl reference solution for 5, 10, 15, 20, 25, 30, 40, 50, or 60 minutes, or longer, while baseline signals are continuously generated. In this embodiment, sensitivity m calibration values can be obtained intermittently by exposing the sensor to an analyte-containing reference solution intermittently, such as but not limited to every 5, 10, 15, 20, 25, 30, 40, 50, or 60 minutes, or longer, such as at the conclusion of the exposure to the 0 mg/dl solution. In other embodiments, such calibrations measurements (e.g., to obtain a baseline value) are performed "intermittently," such as once every 24 hours.

**[0763]** In some embodiments, sensitivity *m* calibration values are obtained substantially continuously/continually, such as by exposure of the sensor to one or more analyte-containing reference solutions every 1, 2, 3, 4, 5, 10, 20, 30, 40, 60 or more minutes. In some embodiments, sensitivity *m* calibration values are obtained substantially continuously/continually by exposure of the sensor to one or more analyte-containing reference solutions for a period of time, such as but not limited to for 5, 10, 15, 20, 25, 30, 40, 50, or 60 minutes, or longer, while sensitivity signals are continuously generated. In one exemplary embodiment, continuous calibration measurements are performed by passing a 100-mg/dl glucose calibration solution across the sensor (e.g., to detect shifts in sensitivity *m*), and the intermittent calibration measurements are performed to determine baseline *b*, by passing a 0-mg/dl calibration solution across the sensor. While not wishing to be bound by theory, it is believed that in some circumstances, baseline drift is greater than sensitivity drift. Accordingly, in some embodiments, the system is configured to perform baseline calibration measurements (with 0-mg/dl glucose) automatically (e.g., every 5-minutes) and a sensitivity calibration measurement (with 100-mg/dl) intermittently (e.g., every hour).

Step Two: Sample Collection and Measurement

**[0764]** In general, the system is configured to allow a sample (e.g., blood) to contact the sensor using the flow control device. Referring now to the top of **Fig. 8B,** the flow control device **604** is configured to draw back (or take-in) a sample (e.g., blood) from the host. For example, to collect a sample, the flow control device **604** reverses and moves backward (e.g., away from the host/catheter), from position **812** to position **810,** thereby causing the pinch point **808** to move away from the host. As the pinch point is moved from position **812** to position **810,** the tube **606** (on the host side of the pinch point **808)** expands (e.g., the tube volume increases).

**[0765]** Referring now to the bottom of **Fig. 8B**, as the tube volume increases, a small, temporary vacuum is created, causing sample **814** (e.g., blood) to be taken up into the catheter lumen **12a.** In some embodiments, the flow control device **604** is configured to take up a sufficient volume of sample **814** such that at least the sensor's electroactive surfaces are contacted by the sample **814.** In some embodiments, a sample volume of from about 1 μl or less to about 2 ml or more is taken up into the catheter **12** and is sufficient to cover at least the electroactive surfaces of the sensor **14.** In some preferred embodiments, the sample volume is from about 10 μl to about 1 ml. In some preferred embodiments, the sample volume is from about 20 μl to about 500 μl. In other preferred embodiments, the sample volume is from about 25 μl to about 150 μl. In more preferred embodiments, the sample volume is from about 2 μl to about 15 μl.

**[0766]** In some embodiments, the sample taken up into the catheter is taken up substantially no farther than the skin (or a plane defined by the skin of the patient). In some embodiments, the sample is taken up into the catheter substantially no farther than the catheter's inner lumen (e.g., substantially not into the IV tubing.)

**[0767]** In some embodiments, the rate of sample take-up is sufficiently slow that the temperature of the sample substantially equilibrates with the temperature of the surrounding tissue. Additionally, in some embodiments, the rate of sample take-up is sufficiently slow such that substantially no mixing of the sample **814** and solution **602a** occurs. In some embodiments, the flow rate is from about 0.001 ml/min or less to about 2.0 ml/min or more. In preferred embodiments, the flow rate is from about 0.01 ml/min to about 1.0 ml/min. In one exemplary preferred embodiment, the flow rate is from about 0.02 ml/min to about 0.35 ml/min. In another exemplary preferred embodiment, the flow rate is from about 0.0.02 ml/min to about 0.2 ml/min. In yet another exemplary preferred embodiment, the flow rate is from about 0.085 ml/min to about 0.2 ml/min.

**[0768]** As described above, in some embodiments, the system is configured such that the speed of the movement between the first and second discreet positions is regulated or metered to control the flow rate of the fluid through the catheter. In some embodiments, the system is configured such that the time of movement between the first and second discreet positions is from about 0.25 to 30 seconds, preferably from about 0.5 to 10 seconds. In some embodiments, the system is configured such that the time of movement between the first and second discreet positions is from about 0.25 to 30 seconds, preferably from about 0.5 to 10 seconds. In some embodiments, the system is configured such that an amount of pinch of the tubing regulates the flow rate of the fluid through the catheter. In some embodiments, regulate

the fluid flow through a combination of metering and/or pinching techniques, for example. Depending on the type of flow control device (e.g., valve), a variety of methods of metering and/or regulating the flow rate can be implemented as is appreciated by one skilled in the art.

**[0769]** Measurements of sample analyte concentration can be taken while the electroactive surfaces are in contact with the sample **814.** An electronics module included in the local and/or remote analyzer **608, 610** controls sample analyte measurement, as described elsewhere herein. In some embodiments, one sample measurement is taken. In some embodiments, a plurality of sample measurements are taken, such as from about 2 to about 50 or more measurements and/or at a sample rate of between about 1 measurement per second and about 1 measurement per minute. In some embodiments, the rate is from about 1 measurement per 2 seconds to about 1 measurement per 30 seconds. In preferred embodiments, sample measurements are taken substantially continuously, such as but not limited to substantially intermittently, as described elsewhere herein.

Optional Step: Flush

**[0770]** In some exemplary embodiments, the flow control device **604** is configured to perform one or more steps, in addition to steps one and two, described above. A flush step, during which the sensor **14** and/or catheter **12** are substantially washed and/or cleaned of host sample, is one such optional step.

**[0771]** Referring now to the top of **Fig. 8C,** the exemplary flow control device **604** performs a flush step by moving forward from position **810** (e.g., toward the host/catheter), past position **812** (e.g., around and over the top of structure **804)** and back to position **810.** For convenience, the movement illustrated by an arrow in the top of **Fig. 8C** is referred to herein as the "flush movement."

**[0772]** Referring now to the bottom of **Fig. 8C,** the flush movement pushes forward a volume of solution **602a** (e.g., a third volume) that pushes the collected blood sample **814** into the host. In some embodiments, the third volume of solution **602a** is substantially equal to the first and second volumes described above. In some embodiments, the flush movement is repeated at least one time. In some embodiments, the flush movement is repeated two, three or more times. With the exception of the first flush movement, which pushes the sample **814** back into the host, each repeat of the flush movement pushes a volume of solution **602a** into the host, for example. In some embodiments, the flush movement pushes the third volume of solution **602a** into the host at a rate of from about 0.25 μl/min or less to about 10.0 ml/min or more. In preferred embodiments the flush movement pushes the third volume of solution into the host at a rate of from about 1.0 μl/min to about 1.0 ml/min. In alternative embodiments, the flow control device **604** is moved to a fully opened position (e.g., no pinch) and the flow regulator **602b** is set at a setting that allows more solution (e.g., an increased volume and/or at a faster rate) to infuse into the host than during the calibration phase (e.g., step one, above). In preferred embodiments, the flush movement washes enough blood off of the analyte sensor's electroactive surfaces that the sensor **14** can measure the solution **602a** substantially without any interference by any remaining blood. In some embodiments, the flush step is incorporated into step one, above.

**[0773]** Generally, the solution **602a** is flushed through the catheter **12,** to ensure that a sufficient amount of the sample has been removed from the sensor **14** and the catheter lumen **12a,** such that a calibration measurement can be taken. However, in some embodiments, sample is collected, measured and flushed out, followed by collection of the next sample, substantially without sensor calibration; the flush step can be executed between samples to ensure that the sample being analyzed is substantially uncontaminated by the previous sample. In some embodiments, a relatively extended flush is used, while in other embodiments the flush is just long enough to ensure no blood remains.

**[0774]** In some embodiments, the effectiveness of the flushing movement is dependent upon the solution **602a** composition (e.g., concentrations of sodium chloride, glucose/dextrose, anticoagulant, etc.). Accordingly, the amount of solution **602a** required to ensure that substantially no sample remains in the catheter **12** and/or on the sensor **14** can depend on the solution **602a** composition. For example, relatively more flush movements may be required to completely remove all of the sample when a non-heparinized solution is selected than when a heparinized solution is selected. In some embodiments, the effectiveness of the flushing movement is also dependent upon the flush flow rate. For example, a relatively faster flow rate can be more effective in removing sample from the sensor than a slower flow rate, while a slower flow rate can more effectively move a larger volume of fluid. Accordingly, in some embodiments, the number of flush movements selected is dependent upon the calibration solution and flow rate selected. In some embodiments, the flush step flow rate is from about 0.25 μl/min or less to about 10.0 ml/min or more, and last for from about 10 seconds or less to about 3 minutes or more. In one exemplary embodiment, about 0.33 ml of solution **602a** is flushed at a rate of about 1.0ml/min, which takes about 20 seconds.

**[0775]** In some embodiments, the flush step returns the sample **814** (e.g., blood) to the host, such that the host experiences substantially no net sample loss. Further more, the flush movement washes the sensor **14** and catheter lumen **12a** of a sufficient amount of sample, such that an accurate calibration measurement (e.g., of undiluted solution **602a)** can be taken during the next step of integrated sensor system **600** operations. In some embodiments, the number of sequential flush movements is sufficient to only wash substantially the sample from the sensor **14** and catheter lumen

**12a.** In other embodiments, the number of sequential flush movements can be extended past the number of flush movements required to remove the sample from the sensor and catheter lumen, such as to provide additional fluid to the host, for example.

**[0776]** At the completion of the flush step, the flow control device **604** returns to step one, illustrated in Fig. 8A. In some embodiments, the steps illustrated in **Figs. 8A** through **8C** are repeated, until the system **600** is disconnected from the catheter/sensor, either temporarily (e.g., to move a host to an alternate location for a procedure) or permanently (e.g., at patient discharge or expiration of sensor life time). In some embodiments, additional optional steps can be performed.

Optional Step: Keep Vein Open (KVO)

**[0777]** Thrombosis and catheter occlusion are known problems encountered during use of an IV system, such as when the fluid flow is stopped for a period of time or flows at a too slow rate. For example, thrombi in, on and/or around the catheter **12,** such as at the catheter's orifice **12b** can cause an occlusion. Occlusion of the catheter can require insertion of a new catheter in another location. It is known that a slow flow of IV solution (e.g., saline or calibration fluid; with or without heparin) can prevent catheter occlusion due to thrombosis. This procedure is know as keep vein open (KVO).

**[0778]** In general, to infuse a fluid into a host, the infusion device must overcome the host's venous and/or arterial pressure. For example, during infusion of a hydration fluid, the IV bag is raised to a height such that the head pressure (from the IV bag) overcomes the venous pressure and the fluid flows into the host. If the head pressure is too low, some blood can flow out of the body and in to the tubing and/or bag. This sometimes occurs when the host stands up or raises his arm, which increases the venous pressure relative to the head pressure. This problem can be encountered with any fluid infusion device and can be overcome with a KVO procedure. KVO can maintain sufficient pressure to overcome the host's venous pressure and prevent "back flow" of blood into the tubing and/or reservoir.

**[0779]** In some embodiments, the flow control device **604** can be configured to perform a KVO step, wherein the fluid flow rate is reduced (but not completely stopped) relative to the calibration and/or wash flow rates. In preferred embodiments, the KVO flow rate is sufficient to prevent the catheter 12 from clotting off and is relatively lower than the flow rate used in step one (above). In preferred embodiments, the KVO flow rate is sufficient to overcome the host vessel pressure (e.g., venous pressure, arterial pressure) and is relatively lower than the flow rate used in step one (above). In some embodiments, the KVO flow rate is from about 1.0 μl/min or less to about 1.0 ml/min or more. In some preferred embodiments, the KVO flow rate is from about 0.02 to about 0.2 ml/min. In some more preferred embodiments, the KVO flow rate is from about 0.05 ml/min to about 0.1 ml/min). In some embodiments, the KVO flow rate is less than about 60%, 50%, 40%, 30%, 20%, or 10% of the calibration and/or flush flow rate(s). In some embodiments, the KVO step is performed for from about 0.25 minutes or less to about 20 minutes or more. In preferred embodiments, the solution **602a** flows at a rate such that the temperature of the solution **602a** substantially equilibrates with the temperature of the tissue surrounding the *in vivo* portion of the catheter **12.** Advantageously, equilibrating the solution **602a** temperature with that of the surrounding tissue reduces the effect of temperature on sensor **14** calibration and/or sample measurement, thereby improving sensor accuracy and consistency. In some embodiments, the KVO step can be incorporated into one or more of the flow control device steps of operation described elsewhere herein, including steps one and two, and the flush step, above.

**[0780]** The KVO step can be executed in one or more ways. In some embodiments, the flow control device **604** can be configured to move to at least one addition position, wherein the tube **606** is partially pinched. For example, the flow control device **604** is configured to move to a position such that the pinch point **808** is partially closed/open. For example, in the embodiment shown in **Figs. 8A** through **8C**, the flow control device **604** can be moved forward somewhat past position **812,** such that the roller **802** causes the tube **606** to be partially pinched. In another example, the flow control device **604** can be moved backwards somewhat behind position **810,** such that the roller **802** again causes the tube **606** to be partially pinched. In preferred embodiment, the amount of pinch can be adjusted such that the desired KVO flow rate can be achieved. In some alternative embodiments, KVO is performed by moving the flow control device between positions **810** and **812** (e.g., see **Fig. 8A**) at a reduced speed, such that the flow rate is from about 0.1 μl/min or less to about 0.5 ml/min or more. In some embodiments, the system is configured such that the time of movement between the first and second discreet positions is from about 0.25 to 30 seconds, preferably from about 5 to 15 seconds. In some preferred embodiments, the tubing is pinched fully closed (e.g., between structures **802** and **806)** during the movement from position **810** and **812** (e.g., see **Fig. 8A).** In some preferred embodiments, after the flow control device reaches position **812,** the flow control device flips over the top and back to position **810** (e.g., see **Fig. 8C**) at a substantially rapid speed that the flow rate remains substantially unchanged. In an even further embodiment, during the KVO step the flow control device alternates between the slow and fast movements at least two times, such that the KVO step lasts a period of time.

**[0781]** In some circumstances, signal artifacts can occur due to the location a catheter is implanted and if the host

has moved his arm (where the catheter is implanted/inserted) to certain positions (e.g., holding his arm up, vertically and/or hanging down). While not wishing to be bound by theory, it is believed that these signal artifacts can arise because femoral veins are relatively small (e.g., 1-2 mm diameter), and in some circumstances an inserted catheter can block the flow of at least some incoming blood, such that the incoming blood is "diverted" around the implantation site by flowing through adjacent alternative veins and/or capillaries. As a result, the blood in the vein containing the catheter can be diluted for a period of time (e.g., after flushing), which can lead to diluted analyte values, which appear as signal artifacts. As time passes, the dilution of the sample by the saline flush dissipates and undiluted blood samples are collected, which leads to termination of the signal artifact(s). The way the host holds his arm can affect the length of time required for the signal artifact to dissipate. For example, in some embodiments, if the host holds his arm at chest level, the vein is filled by blood at a first rate, and the sample dilution is dissipated within a first time period. If the host holds his arm down low, the blood flows through the vein at a second rate that is faster than the first rate, and the dilution is relieved sooner (e.g., within a second time period that is shorter than the first time period). Conversely, if the host raises his arm over his head, the blood flows into the vein at a third rate that is slower than the first rate; which results in the dilution dissipating within a third period of time that is longer than the first period of time.

**[0782]** In some embodiments, the signal artifacts resulting from blocking of the vein at the site of catheter implantation and subsequent sample dilution can be substantially eliminated (or reduced/shortened) by reducing the volume of 0 mg/dl solution used to wash away the 100-mg/dl calibration solution, depending upon the volume of the 100 mg/dl calibration solution used to calibrate the sensor. For example, in some embodiments a 0.5X, 1X, 2X, 3X, 4X, 5X, 6X, 7X, 8X, 9X, 10X (or greater) volume of the 0-mg/dl solution can be used to sufficiently wash out the volume 100 mg/dl calibration solution. In some embodiments, wherein a sufficiently small volume of 100 mg/dl solution is used during calibration, the flushing can be done at the KVO rate. In some embodiments, signal artifacts are prevented by using smaller volumes of calibration/wash fluids but switching more frequently between the fluids.

Maintaining Patency During A Sensor Session

**[0783]** In some embodiments, the analyte sensor is implanted in the host (e.g., *via* a vascular access device) for an extended period of time. For example, in some embodiments, a sensor session can last 3, 5, 7, 10, 21, 30 or more days. As used herein, the term "sensor session" is a broad term and refers without limitation to the period of time of sensor is applied to (e.g., implanted in) the host and is being used to obtain sensor values. For example, in some embodiments, a sensor session extends from the time of sensor (e.g., including implanting the vascular access device) implantation to when the sensor is removed. During this period of time, the vein's condition can deteriorate, such that vein and/or vascular access device is no longer patent (e.g., freely open, not occluded), and the system can no longer function optimally. While not wishing to be bound by theory, it is believed that patency can be substantially maintained during a sensor session by metering a reference/calibration solution through the vascular access device a sufficient amount of time (e.g., a percentage of the duration of the sensor session). As used herein, the phrase "a sufficient amount" is a broad term and refers without limitation to an amount that provides a desired function. For example, a sufficient amount can be a sufficient amount of time, a sufficient amount of fluid volume, and the like. In some embodiments, a sufficient amount can be expressed numerically, such as a percent (%), a volume, a weight, a period of time (e.g., minutes, hours, days, months), and the like. For example, in some embodiments, the flow control device is configured to meter a sufficient amount of a reference solution (e.g., through the vascular access device) such that the analyte sensor contacts the reference solution at least about 50%, 55%, 60%, 65%, 70%, 75%, 80%, 85%, 90% or 95% of the time during a sensor session. It is generally preferred that the analyte sensor contacts the reference solution from about 50% to about 80%, 85%, 90% or 95% of the time during a sensor session In some embodiments, the sensor is located in or on the vascular access device and the flow control device meters the reference solution through the vascular access device for a sufficient amount of time (e.g., a portion of the sensor session), with a sufficient flow rate (e.g., from about 0.05 ml/min to about 0.5 ml/min, preferably about 0.1 ml/min) that the vascular access device remains patent during a sensor session. Advantageously, the flow rate is sufficient to maintain a patent vessel without infusing excess fluid. In a preferred embodiment, the vascular access device remains patent during a sensor session of at least about 1, 3, 5, 7, 10, 15, 20, 25, or 30 days, or longer. In one exemplary embodiment, the flow control device is configured to meter the reference solution through the vascular access device for at least about 50% of a sensor session, at a flow rate from about 0.001 ml/min to about 2.0 ml/min, such that the vascular access device remains patent during a sensor session of at least about 5 days. In another exemplary embodiment, the flow control device is configured to meter the reference solution through the vascular access device for at least about 65% of a sensor session, at a flow rate from about 0.001 ml/min to about 2.0 ml/min, such that the vascular access device remains patent during a sensor session of between about 5 days and about 30 days. In still another exemplary embodiment, the flow control device is configured to meter the reference solution through the vascular access device for between about 50% and about 80% of a sensor session, at a flow rate from about 0.001 ml/min to about 2.0 ml/min, such that the vascular access device remains patent during a sensor session of at least about 30 days.

Preventing Sensor Biofouling During A Sensor Session

**[0784]** As discussed above, sensor sessions can last from about 3 days to 30 or more days. During this period of time, the sensor is repeatedly exposed to (contacted with) a bodily fluid (e.g., blood). In some circumstances, during sensor exposure to blood, some blood components/material, such as but not limited to proteins, lipids, carbohydrates and cells, can "stick" to the sensor, such that a layer of this material coats at least part of the sensor and the sensor can no longer function accurately. This process of blood components sticking to the sensor and disrupting the sensor's function is generally referred to as "biofouling." While not wishing to be bound by theory, it is believed that biofouling can be substantially reduced and/or eliminated by limiting the length of time the sensor is exposed to blood and/or by maintaining the sensor in a reference solution (or saline) a substantial portion of the sensor session, whereby sensor accuracy is maintained throughout the sensor session. Washing the sensor is described in detail in the section entitled "Maintaining Patency During A Sensor Session."

**[0785]** As a non-limiting example, in some embodiments, the flow control device is configured to meter the reference solution, such that the reference solution contacts the sensor a substantial amount of time such that biofouling does not occur for at least about 3 days of sensor use. For example, the sensor can be contacted with the reference solution about 50%, 60%, 70%, 80%, 90%, or 95% of the sensor session duration. In preferred embodiments, the system is configured such that biofouling does not occur for at least about 7, 10, 21, 30 or more days of sensor use.

**[0786]** In one exemplary embodiment, wherein the sensor is located in or on a vascular access device, the system includes a flow control device configured to meter a reference solution through the vascular access device for a sufficient amount of time with a sufficient flow rate that the vascular access device remains patent during a sensor session of at least about 3 days. For example, in some embodiments, the flow control device meters the reference solution through the vascular access device for at least about 50% of a sensor session, at a flow rate from about 0.001 ml/min to about 2.0 ml/min, such that the vascular access device remains patent during a sensor session of at least about 5 days. In another exemplary embodiment, the flow control device meters the reference solution through the vascular access device for at least about 65% of a sensor session, at a flow rate from about 0.001 ml/min to about 2.0 ml/min, such that the vascular access device remains patent during a sensor session of between about 5 days and about 30 days. In still another exemplary embodiment, the flow control device meters the reference solution through the vascular access device for between about 50% and about 80% of a sensor session, at a flow rate from about 0.001 ml/min to about 2.0 ml/min, such that the vascular access device remains patent during a sensor session of at least about 30 days.

Alternative Flow Control Device Configurations

**[0787]** As disclosed above, the flow control device **604** can be configured a variety of ways, which can require modifications to one or more of the steps of operation described above. For example, in some embodiments, the flow control device **604** can be configured to include a simple pinch valve, wherein the valve can be configured to open, close or partially open. In some embodiments, the flow control device **604** can be configured to include a non-linear rolling pinch valve, wherein the roller can move back and forth between opened, closed and partially opened positions, for example.

**[0788]** In some embodiments, the flow control device **604** can include one roller **802** (e.g., first structure) attached to an axle **804** and configured to press against a curved surface **806** (e.g., second structure), such that when the roller **802** is pressing against the curved surface **806** at or between positions **810** and **812,** the tubing **606** is pinched completely closed and the flow control device **604** moves the roller **802** forward (e.g., toward the host). In one exemplary embodiment, the flow control device **604** can be configured to perform step one (above, contacting the sensor **14** with solution **602a)** by moving the roller **802** forward (e.g., rotating from position **810** to **812,** see **Fig. 8A),** thereby causing solution **602a** to flow over the sensor **14.** In some embodiments, the flow control device **604** is configured to perform step two (contacting the sensor **14** with sample) by moving the roller **802** backwards (e.g., rotating from position **812** to **810,** see **Fig. 8B),** causing blood **814** to enter the catheter **12** and contact the sensor **14.** Additionally, the flow control device **604** can be configured to perform a wash or KVO step by moving the roller **802** forward (from position **810)** past position **812** and around the axle **804** until position **810** is again reached a plurality of times sequentially (e.g., see **Fig. 8C).** In a further example, the flow control device **604** includes two, three or more rollers **802** arranged about axle **804.** In some embodiments, the flow control device includes a plurality of rollers arranged about the axle, wherein the flow control device performs KVO by rotating the rollers about the axle a plurality of times, to continuously push (e.g., for a period of time) the solution forward into the host.

**[0789]** In one alternative embodiment, back flow can be substantially stopped by incorporation of a one-way, pressure-controlled valve into the system, such as at or adjacent to the catheter or sensor connector, whereby fluid can flow into the host only when fluid pressure (e.g., head pressure) is applied to the reservoir-side of the valve. In other words, fluid can only flow in the direction of the host (e.g., toward the host), not backwards towards the reservoir. In some embodiments, the valve is a two-way valve configured such that the pressure required to open the valve is greater than the venous pressure, such that back flow is substantially prevented.

**[0790]** The preferred embodiments provide several advantages over prior art devices. Advantageously, the movement of the solution **602a** and sample occur at a metered rate and are unaffected by changes in head pressure, such as but not limited to when the host elevates his arm or gets up to move around. Also, sample loss to the host is minimized, first by returning all collected samples to the host; and second by substantially preventing back-flow from the host (e.g., into the tubing or reservoir) with a "hard stop" (e.g., a point beyond which the flow control device cannot move fluid into or out of the host). For example, in one preferred embodiment, the flow control device can be configured to deliver no more than 25-ml of solution to the host per hour. In another exemplary embodiment, the flow control device can be configured to draw back no more than 100 μl of blood at any time. Advantageously, the flow rate of solution **602a** and sample **814** is carefully controlled, such that both the sample **814** and the solution **602a** remain substantially undiluted. Additionally, the solution **602a** warms to the host's local body temperature, such that the integrated sensor system **600** is substantially unaffected by temperature coefficient and sensor **14** accuracy is increased.

Pumpless Sample Withdrawal

**[0791]** In some circumstances, it is preferred to meter the flow of a fluid through a vascular access device (including withdrawal of at blood sample) without the use of a pump and/or a flow control device (e.g., described above). Accordingly, some embodiments provide a system for continuously measuring an analyte in an artery of a host *in vivo,* which does not require the use of the flow control device of the preferred embodiments or of a pump. Accordingly, in some preferred embodiments, the system includes an arterial infusion system and continuous analyte sensor coupled thereto. The arterial infusion system is configured and arranged to meter the flow of a fluid into and/or out of an artery of a host, and includes an arterial catheter, a pressure transducer, an infusion fluid, and a pressure system. The pressure system is configured to increase and/or reduce an amount of pressure applied to the infusion fluid, such that when the infusion system is applied to the host (e.g., the catheter is implanted in the host's artery), the pressure system can infuse the infusion fluid, withdraw a blood sample, and reinfuse a withdrawn sample into the host. In general, arteries are pressurized. Accordingly, blood will expelled from a puncture in the artery (e.g., an inserted/implanted catheter, a cut or breakage) unless pressure greater than the arterial pressure is applied thereto, such as via compression, a pressure cuff, a pressurized infusion system, and the like. An arterial pressure system can be configured to infuse fluid into the host by increasing the pressure applied to the infusion fluid, such as (but not limited to) by increasing the pressure applied with a blood pressure cuff, such that the applied pressure overcomes the arterial pressure. In preferred embodiments, the system is configured to withdraw a sample (e.g., contact the sensor with the blood sample) by reducing the applied pressure (in a controlled manner) until a sample of blood is pushed into the catheter (by the arterial pressure) and contacts the sensor. In some embodiments, the withdrawn sample is reinfused into the host, such as by increasing the applied pressure such that the arterial pressure is again overcome and infusion fluid flows into the host. In some embodiments, the withdrawn sample is diverted to waste disposal, such as *via* a valve. In some embodiments, the system is configured such that the sensing portion of the sensor are disposed within the host's body (e.g., within the artery) as described herein. However, in other embodiments, the sensor is disposed extracorporeally (e.g., above a plane defined by the host's skin) and the sample is withdrawn out of the host's body. In preferred embodiments, the system includes electronics configured to regulate the pressure system, such that infusion and sample withdrawal are controlled.

**[0792]** The analyte sensor can be configured to detect a variety of analytes, as described elsewhere herein. In some embodiments, the analyte sensor includes a single working electrode, which generates a first signal associated with the concentration of the analyte in the sample. In other embodiments, the analyte sensor is a dual-electrode continuous analyte sensor, as described elsewhere herein, and includes a first working electrode configured to generate the first signal (comprising an analyte-related signal component and a non-analyte-related signal component) and a second working electrode is configured to generate a second signal (comprising the non-analyte related signal component).

**[0793]** In some embodiments, a method for continuously measuring an analyte in an artery of a host *in vivo* is provided. In this embodiment, the method includes the steps of coupling a continuous analyte sensor with an arterial catheter system applied to a host, wherein the sensor is configured to generate an analyte-related signal associated with an analyte in a sample, and wherein the arterial catheter system comprises an arterial catheter, a pressure transducer, an infusion fluid, and a pressure system configured to increase and/or reduce an amount of pressure applied to the infusion fluid; reducing the amount of pressure, such that a sample of arterial blood contacts the sensor; and generating the analyte-related signal with the sensor. In some embodiments, the coupling step comprises coupling the sensor to the arterial catheter, such as by inserting the sensor into a lumen of the arterial catheter. In some embodiments, the method includes a step of reinfusing the sample into the host, such as by increasing the amount of pressure. In some embodiments, the arterial blood pressure of the host is monitored, using the pressure transducer. Arterial pressure monitors are known in the art. In preferred embodiments, the analyte-related signal is processed to provide an analyte value. In some embodiments, the signal is also calibrated.

**[0794]** In some embodiments, the generating step further includes generating a second signal with the sensor, wherein sensor comprises a first working electrode configured to generate a first signal comprising an analyte-related signal

component and a non-analyte-related signal component and the second working electrode is configured to generate the second signal comprising the non-analyte-related signal component. The first and second signals can be processed, to provide a processed signal substantially without a signal component due to the non-analyte-related signal component, and/or to provide a scaling factor. In some embodiments, the generating step further comprises generating a reference signal associated with a reference analyte in the sample, wherein the sensor further comprises a reference sensor configured to generate the reference signal.

Systems and Methods for Processing Sensor Data

**[0795]** In general, systems and methods for processing sensor data associated with the preferred embodiments and related sensor technologies include at least three steps: initialization, calibration, and measurement. Although some exemplary glucose sensors are described in detail herein, the systems and methods for processing sensor data can be implemented with a variety of analyte sensors utilizing a variety of measurement technologies including enzymatic, chemical, physical, electrochemical, spectrophotometric, polarimetric, calorimetric, radiometric, and the like. Namely, analyte sensors using any known method, including invasive, minimally invasive, and non-invasive sensing techniques, configured to produce a data signal indicative of an analyte concentration in a host during exposure of the sensor to a biological sample, can be substituted for the exemplary analyte sensor described herein.

**[0796]** In some embodiments, the sensor system is initialized, wherein initialization includes application of the sensor and/or sensor system in or on the host. In some embodiments, the sensor system includes a computer system including programming configured for performing one or more of the following functions: turning the system on, requesting and/or receiving initial data (e.g., time, location, codes, etc), requesting and/or receiving patient data (e.g., age, conditions, medications, insulin dosing, etc), requesting and/or receiving calibration information (e.g., manufacturer calibration lot data, reference information such as solution(s) provided for calibration, etc.), and the like.

**[0797]** In some embodiments, the sensor system is configured with a predetermined initial break-in time. In some embodiments, the sensor's sensitivity (e.g., sensor signal strength with respect to analyte concentration) and/or baseline can be used to determine the stability of the sensor; for example, amplitude and/or variability of sensor sensitivity and/or baseline may be evaluated to determine the stability of the sensor signal. In alternative embodiments, detection of pH levels, oxygen, hypochlorite, interfering species (e.g., ascorbate, urea, and acetaminophen), correlation between sensor and reference values (e.g., R-value), and the like may be used to determine the stability of the sensor. In some embodiments, the sensor is configured to calibrate during sensor break-in, thereby enabling measurement of the biological sample prior to completion of sensor break-in.

**[0798]** In one embodiment, systems and methods are configured to process calibrated sensor data during sensor break-in. In general, signals associated with a calibration and/or measurement phase of the sensor system can be measured during initial sensor break-in. Using a rate method for measuring an analyte (e.g., measuring the rate of change of a step change), a sensor signal can be calibrated with a correction factor to account for the rate of change of the break-in curve. In one exemplary embodiment, the bottom of sequential step responses (e.g., of calibration phases during sensor break-in) can be fit to a line or curve (e.g., using linear or non-linear regression, such as least squares regression), to extrapolate the rate of change of the curve of the sensor break-in. Accordingly, the rate of change measured in a measurement phase can be corrected to account for the rate of change of the sensor break-in curve, and the sensor signal calibrated. By calibrating during sensor break-in, sensor data can more quickly be provided (e.g., to the user interface) after sensor insertion.

**[0799]** In some embodiments, systems and methods are configured to determine an initial baseline value of the sensor. In general, baseline refers to a component of an analyte sensor signal that is not substantially related to the analyte concentration In one example of a glucose sensor, the baseline is composed substantially of signal contribution due to factors other than glucose (for example, interfering species, non-reaction-related hydrogen peroxide, or other electroactive species with an oxidation/reduction potential that overlaps with hydrogen peroxide).

**[0800]** In preferred embodiments, the sensor system includes a computer system including programming configured to determine calibration information and calibrate a signal associated with a biological sample there from. In general, calibration of the signal includes initial calibration, update calibration and/or re-calibration of the sensor signal. Although some systems and methods for calibrating a sensor are described in more detail elsewhere herein, for example in the section entitled, "Sensor Electronics," additional and alternative methods for providing calibration information and calibrating the sensor's signal are provided in the following description and can be used in combination with and/or alternative to the methods described elsewhere herein.

**[0801]** The term "calibration information" generally refers to any information, such as data from an internal or external source, which provides at least a portion of the information necessary to calibrate a sensor. In some embodiments, calibration information includes steady state information, such as baseline information and/or sensitivity information obtained by processing reference data from an internal and/or external reference source, which is described in more detail elsewhere herein. In some embodiments, calibration information includes transient information, such as rate of

change information and/or impulse response information obtained by processing a signal produced during exposure of the sensor to a step change (e.g., sudden or nearly sudden change) in analyte concentration, which is described in more detail elsewhere herein.

**[0802]** In some embodiments, steady state information includes reference data from an external source, such as an analyte sensor other than the sensor of the sensor system configured to continuously measure the biological sample, also referred to as external reference data or external reference value(s). In some embodiments, calibration information includes one, two, or more external reference values (e.g., from self-monitoring blood glucose meters (finger stick meters), YSI Glucose Analyzer, Beckman Glucose Analyzer, other continuous glucose sensors, and the like). In some embodiments, one or more external reference values are requested and/or required upon initial calibration. In some embodiments, external reference value(s) are requested and/or required for update calibration and/or re-calibration. In some embodiments, external reference values are utilized as calibration information for calibrating the sensor; additional or alternatively, external reference values can be used to confirm the accuracy of the sensor system and/or to detect drifts or shifts in the baseline and/or sensitivity of the sensor.

**[0803]** In one exemplary embodiment, at least one external reference value in combination with at least one internal reference value together provide calibration information useful for calibrating the sensor; for example, sensitivity of a sensor can be determined from an external reference value and baseline can be at least partially determined from an internal reference value (e.g., a data signal indicative of an analyte concentration in a reference solution during exposure of the sensor to the reference solution, which is described in more detail elsewhere herein).

**[0804]** In another exemplary embodiment, calibration information includes two or more external reference values that provide calibration information useful for calibrating the sensor; for example, at least two SMBG meter values can be used to draw a calibration line using linear regression, which is described in more detail elsewhere herein.

**[0805]** In yet another exemplary embodiment an external reference value is utilized to confirm calibration information otherwise determined (e.g., using internal reference values).

**[0806]** In some embodiments, steady state information includes reference data obtained from the analyte sensor to be calibrated, also referred to as internal reference data or internal reference values. In one exemplary embodiment, internal reference data includes a signal associated with exposure of the sensor to one or more reference solutions (e.g., calibration solutions), which is described in more detail elsewhere herein.

**[0807]** In some embodiments, the sensor system includes one or more reference solutions (e.g., calibration solutions in some embodiments), wherein the system is configured to expose the sensor to the one or more reference solution(s) to provide calibration information (e.g., an internal reference value), such as baseline and/or sensitivity information for the sensor. In one exemplary embodiment, a reference solution including a known analyte concentration is provided, wherein the system is configured to expose the sensor to the reference solution, and wherein the system is configured to produce a data signal indicative of an analyte concentration in the reference solution during exposure of the sensor to the reference solution, as described in more detail elsewhere herein. In some embodiments, two reference solutions with two different analyte concentrations are provided. For example, in order to generate reference values for detecting sensitivity drift of a glucose sensor, two saline solutions containing 100 and 200 mg/dl glucose respectively can be provided.

**[0808]** In general the system can be configured to obtain internal reference values at one or more time points, intermittently, and/or continuously. For example, in some embodiments, calibration for drift in baseline and/or sensitivity can be done at set time intervals, depending upon the severity of the drift. In some circumstances, it is preferred to calibrate very frequently (e.g., between about every 1 minute or less and about every 2, 3, 4, 5, 10, 15, 20 or 30 minutes or longer). In other circumstances, it is preferred to calibrate less frequently (e.g., about every 1, 2, 3, 5, 10 15 or 24 hours or longer). For example, in some circumstances, baseline drift has a substantial effect on sensor accuracy, while sensitivity drift has little effect. Accordingly, a baseline calibration solution (e.g., 0-mg/dl glucose) can be used to calibrate the baseline about every 5 minutes. Thus, to calibrate for sensitivity drift, an analyte-containing calibration solution (e.g., 100-mg/dl glucose in saline) can be used to calibrate the sensor less frequently, such as about once every 1, 2, 3, 5, 10, 12, 24, 48 or more hours. In some embodiments, one or more external reference values, such as reference values obtained by testing a blood sample with SMBG or a YSI device, can be used to calibrate the system, in addition to the internally provided reference values (e.g., provided *via* the calibration solutions).

**[0809]** Although much of the description focuses on the use of a reference calibration solution to provide an internal reference value, other sensor technologies, such as optical sensing methods, are known to provide one or more internal reference standards (e.g., of known absorbance, reflectance, fluorescence, etc) to determine baseline and/or sensitivity information, as is appreciated by one skilled in the art; accordingly, the systems and methods described herein can be implemented with other types of internal reference values. Examples of analyte sensors configured for optical detection of the analyte and/or a reference analyte are described in detail in the section entitled "Optical Detection," above. In some embodiments, a "plateau" is reached when the sensor has been exposed to the sample of bodily fluid (e.g., blood) or a reference solution a sufficiently long period of time that the sensor's enzyme has used up (e.g., reacted with, detected) substantially all of the available analyte.

**[0810]** In some embodiments, the sensor system is configured to use a steady state measurement method, from which steady state information can be obtained. Steady state information can be obtained during exposure of the sensor to an analyte concentration when the signal has reached a "plateau" wherein the signal is representative of the analyte concentration; the term plateau does not limit the signal to a flat signal, rather the plateau represents a time point or time period during which the signal is substantially stable and a data point that represents the analyte concentration can be reliably obtained.

**[0811]** **Fig. 10** is a graph that schematically illustrates a signal produced during exposure of the sensor to a step change in analyte concentration, in one exemplary embodiment. The x-axis represents time; the y-axis represents sensor signal (e.g., in counts). In general, a step change occurs when a sensor is sequentially exposed to first and second different analyte concentrations, wherein the signal (after the change from exposure of the sensor to the first analyte concentration to exposure of the sensor to the second analyte concentration) includes a measurable rate of change (transient information) that subsequently "plateaus" or substantially "plateaus" to a signal that substantially represents the analyte concentration to which the sensor is exposed (steady state information). As one example, a step change occurs when a sensor is exposed to a reference solution of a first analyte concentration and then subsequently exposed to a reference solution of a second, different, analyte concentration. As another example, a step change occurs when a sensor is exposed to a reference solution of a known analyte concentration and then subsequently exposed to a biological sample of unknown or uncalibrated analyte concentration.

**[0812]** Referring to **Fig. 10**, at a first time point **1002,** a sensor is exposed to a step change in analyte concentration, for example, from a zero concentration reference analyte solution to a biological sample of unknown or uncalibrated analyte concentration. During the initial signal response to the step change, a rate of change **1004** of the signal can be measured for a time period. In some embodiments, for example when the step change is between two known reference solutions, the rate of change information can provide transient information useful for calibrating the sensor, which is described in more detail elsewhere herein. However, if either of the first and/or second analyte concentrations of the step response is not known, the rate of change information, alone, cannot provide sufficient calibration information necessary to calibrate the sensor.

**[0813]** Point **1006** represents a point in time that the signal response shifts from transient information (e.g., rate of change) to steady state information (e.g., plateau), in some embodiments. Namely, the signal, beginning at point **1006,** substantially accurately represents the analyte concentration and can be used in steady state equations to determine an analyte concentration, in some embodiments. In one exemplary embodiment of steady state equations useful for calibrating the sensor system, the calibration information is obtained by solving for the equation $y=mx+b$, wherein: "$y$" represents the sensor data value (e.g., digitized in "counts") determined at a single point (or averaged value over a window of data where signal is indicative of analyte concentration, for example); "$b$" represents baseline (e.g., unrelated to the analyte); "$m$" represents sensitivity (e.g., for a glucose sensor, counts/mg/dL); and "$x$" is the concentration of the reference solution (e.g., known analyte concentration in a reference calibration solution (e.g., glucose in mg/dL)). In this exemplary embodiment, steady state information includes sensitivity and baseline.

**[0814]** In some embodiments, the sensor data value (y) can be obtained from a moving window that intelligently selects a plateau during exposure of the sensor to an analyte concentration. In some embodiments, the sensor system is configured to be exposed to two or more known reference calibration solutions from which steady state information (sensitivity and baseline) can be processed to calibrate the sensor system; namely, by providing two known analyte concentrations, the steady state equation described above can be utilized to solve for baseline and sensitivity of the sensor, which can be utilized to define a conversion function or calibration factor, such as described in more detail elsewhere herein.

**[0815]** Referring again to **Fig. 10,** point **1006** is a point that can be used as "$y$" in the steady state equation described above. In some embodiments, the point **1006** is easily determinable as it is the beginning of a signal plateau **1008** (represented by a dashed line); accordingly, the system includes programming to process the data signal to determine the signal plateau and/or a time point therein. In general, a step change produces a signal plateau in the signal response, which is indicative of a steady state response to the analyte concentration measurement. In some embodiments, the system includes programming configured determine the time period (window) during which the signal has reached a plateau and choose a single point or average point from that window.

**[0816]** In some situations, however, the point **1006** and/or plateau **1008** may not be easily determinable. For example, in some sensor systems, the diffusion of certain non-analyte species (e.g., baseline, background and/or interfering species), which may diffuse more slowly than the analyte (e.g., through a membrane system that covers the analyte sensor), do not reach a steady state during the same time period that the analyte reaches a steady state. In these situations, the signal may not "plateau" in a measurable manner because of the reaction of the lagging species through the membrane system, which generate additional signal over the actual analyte plateau **1008**. In other words, while the analyte concentration may have reached a plateau, the baseline has not. Dashed line **1010** represents the signal response to a step change in such a situation, for example, wherein the signal does not substantially "plateau" due to the lagging diffusion of certain non-analyte species, resulting in a non-measurable analyte plateau. In these situations, additional

information is required in order to provide calibrated analyte sensor data. Systems and methods for providing additional information and/or to provide sufficient calibration information to calibrate an analyte sensor in such situations are described in more detail below, with reference to conjunctive measurements, for example.

**[0817]** In some embodiments, the sensor system is exposed to a reference solution with a known analyte concentration of about zero, and wherein the steady state information comprises baseline information about the sensor in the reference solution. For example, a glucose sensor system can be exposed to a 0 mg/dl glucose solution (e.g., an isotonic solution without any glucose concentration) and the signal associated with the zero glucose concentration in the reference solution provides calibration information (steady state) indicative of at least a portion of the baseline of the sensor. However, the signal associated with the zero glucose concentration in a reference solution (such as saline) may not be equivalent to the baseline signal when the sensor is exposed to a biological sample (e.g., blood) from which the sensor is configured to obtain its analyte concentration measurement; accordingly, additional calibration information may be required in order to determine baseline of a biological sample (e.g., blood) in some embodiments. In some embodiments, the calibration solution includes additional components provided to overcome baseline in blood, for example. In some embodiments, a factor can be determined (e.g., from historical data) to determine an adjustment factor for a difference between baseline in the biological sample (e.g., blood) and baseline in the reference solution. In some embodiments, baselines of the working electrodes can be determined prospectively, such as by testing in the reference solution by the manufacturer. In some embodiments, the difference in baseline of a biological sample (e.g., blood) and the baseline of the reference solution, also referred to as $b_{offset}$ herein, can be determined using other techniques, such as described in more detail below.

**[0818]** In general, the calibration information described above, including a known baseline and sensitivity, can be used to determine a conversion function or calibration factor applied to convert sensor data ("*y*") into blood glucose data ("*x*"), as described in more detail elsewhere herein.

**[0819]** In some embodiments, systems and methods are configured to obtain transient measurement information associated with exposure of the sensor to a reference solution of known analyte concentration and/or a biological fluid of unknown or uncalibrated analyte concentration. In some embodiments, the system is configured obtain transient information by exposing the sensor to a step change in analyte concentration and process the rate of change of the associated signal. In some embodiments, the system is configured to obtain transient information by exposing the sensor to a step change in analyte concentration and processing the impulse response of the associated signal.

**[0820]** In one exemplary embodiment, the sensor is exposed to a first reference solution of a known analyte concentration and then to a second reference solution of a known analyte concentration to determine the rate of change of the signal response. In these embodiments, the equation ($\Delta y/\Delta t = r \cdot \Delta x$) can be used to obtain the transient information, wherein "$\Delta x$" is the difference between the two known solutions that are being measured (e.g., 0 mg/dL to 100 mg/dL in an exemplary glucose sensor), "$\Delta y$" is the measured difference between the sensor data (e.g., in counts) corresponding to the analyte concentration difference in known reference solutions ($\Delta x$), "$\Delta t$" is the time between the two "*y*" sensor measurements referenced with $\Delta y$, and "*r*" represents the rate of change calibration factor, or rate of change conversion function, that can be applied for that particular sensor to obtain calibrated blood glucose measurements from sensor rate of change data.

**[0821]** In some embodiments, transient information can be obtained from the rate of change of a signal produced during exposure of the sensor to a biological sample of unknown or uncalibrated analyte concentration. In some embodiments, transient information can be obtained from the step and/or impulse response of a signal produced during exposure of the sensor to a step change in analyte concentration.

**[0822]** In some embodiments, neither steady state information, nor transient calibration measurements are used in isolation in calibrating the sensor system, but rather steady state and transient information are combined to provide calibration information sufficient to calibrate sensor data such as described in more detail, below. For example, in some embodiments, wherein baseline is not completely known (e.g., $b_{offset}$ must be determined), wherein a rate of change calibration factor is not easily determinable (e.g., when multiple known reference solutions cannot be pushed substantially immediately adjacent to each other to provide a rate of change indicative of the step or impulse response), wherein the a steady state measurement cannot be obtained (e.g., due to lagging species affecting the analyte signal plateau), and the like. In some embodiments, both steady state information and transient information are processed by the system to provide sensor calibration, confirmation, and/or diagnostics. In some embodiments, transient sensor information from unknown or uncalibrated blood glucose measurements can be processed to provide calibration information for the sensor system, such as described in more detail below.

**[0823]** In some embodiments, once at least a portion of the calibration information is determined, the sensor system is configured to expose the sensor to a biological sample and measure a signal response thereto. In some embodiments, the sensor can be continuously exposed to the biological sample, wherein at least some external reference values are used as calibration information for calibrating the sensor system. In some embodiments, the sensor can be intermittently exposed to the biological sample, wherein at least some internal reference values are used as calibration information for calibrating the sensor system, also referred to as auto-calibration in some exemplary embodiments.

**[0824]** In some embodiments, the sensor system is calibrated solely using steady state information, such as described in more detail elsewhere herein. In one such embodiment, the sensor system is configured to be exposed to a biological sample and a value (y) determined from the signal plateau, which is used in combination with a conversion function (calibration factor) that uses steady state information (e.g., sensitivity and baseline) to obtain a calibrated analyte concentration (e.g., glucose concentration in mg/dL or mmol/L) equivalent to the measured sensor data value y.

**[0825]** In general, the sensor system of the preferred embodiments can be configured to utilize any combination the steady state information (e.g., from external and/or internal sources) described in more detail elsewhere herein. In some embodiments, the sensor system includes systems and methods configured to calibrate the sensor based on one, two, or more external reference values. In some embodiments, the sensor system includes systems and methods configured to calibrate the sensor based on one or more external reference values, which calibration can be confirmed using an internal reference value (e.g., zero analyte concentration reference solution). In some embodiments, the sensor system includes systems and methods configured to calibrate the sensor based on one external reference value in combination with one internal reference value to determine baseline and sensitivity information. In some embodiments, the sensor system includes systems and methods configured to calibrate the sensor based on internal reference values, also referred to as auto-calibration. In general, auto-calibration includes the use of one or more reference solution to calibrate the sensor system. In some embodiments, the sensor system includes systems and methods configured to calibrate the sensor based on prior information, which is described in more detail elsewhere herein. In some embodiments, the sensor system includes systems and methods configured to calibrate the sensor based on dual working electrodes, by substantially eliminating the baseline component of the steady state calibration equation (e.g., (y=mx)).

**[0826]** In some embodiments, the sensor system includes systems and methods configured to calibrate the sensor based solely on transient information (e.g., rate of change, decay, impulse response, etc) described in more detail elsewhere herein. In one exemplary embodiment, analyte concentration can be determined from the change in sensor data responsive to a step change ($\Delta x$), the time ($\Delta t$) elapsed between the sensor data measurements $\Delta y$, and the rate of change calibration factor/rate of change conversion function, such as described in more detail above.

**[0827]** In some embodiments, the sensor system includes systems and methods configured to calibrate the sensor based on conjunctive information, wherein the calibration information used to calibrate the sensor system includes both steady state information and transient information.

**[0828]** In one exemplary embodiment, the sensor system includes systems and methods configured to calibrate the sensor based on a rate of change (transient information) associated with a signal produced during exposure of the sensor to a step change between a reference solution of known analyte concentration (e.g., 0 mg/dl glucose) and a biological sample; in this exemplary embodiment, a reference value (steady state information) from an external analyte sensor (e.g., blood glucose meter) can be obtained for the analyte concentration in the biological sample, thereby providing sufficient information to solve for calibration using rate of change of the signal response to the step change there between. One advantage of using rate of change calibration methods includes its insensitivity to baseline and interfering species.

**[0829]** In one preferred embodiment, a system is provided for monitoring analyte concentration in a biological sample of a host, the system including: a substantially continuous analyte sensor configured to produce a data signal indicative of an analyte concentration in a host during exposure of the sensor to a biological sample; a reference solution including a known analyte concentration, wherein the system is configured to expose the sensor to the reference solution, and wherein the sensor is configured to produce a data signal indicative of an analyte concentration in the reference solution during exposure of the sensor to the reference solution; and a computer system including programming configured to determine calibration information and calibrate a signal associated with a biological sample there from, wherein the calibration information includes steady state information and transient information. In some embodiments, the calibration information is determined from a signal associated with exposure of the sensor to the reference solution and a signal associated with exposure of the sensor to a biological sample.

**[0830]** One situation wherein steady state information and transient information are useful together for calibrating a sensor system includes a situation where a baseline measurement obtained from an internal reference ($b_{reference}$) provides only a portion of the baseline information necessary for calibrating the sensor system. As one example, the baseline of blood is different from the baseline of saline (e.g., reference) and compounds or molecules that make up the baseline in blood can create artifacts (e.g., $b_{offset}$), which can make calibration using internally derived steady state information alone, difficult. Namely, plateau **1008 (Fig. 10)** in the signal responsive to the step change in analyte concentration does not occur in blood, in some embodiments, due to slow diffusion of baseline-causing compounds/molecules to the sensor electroactive surface; instead, an artifact **1010 (Fig. 10)** is observed in the signal. Accordingly, in some embodiments, baseline information useful for calibration of a sensor system includes both $b_{reference}$ and $b_{offset}$ A variety of systems and methods of determining $b_{offset}$, which can be useful in providing calibration information and/or diagnostics and fail-safes, has been discovered, as described in more detail elsewhere herein.

**[0831]** In some embodiments, $b_{offset}$ can be determined from transient information derived from a signal associated with exposure of the sensor to a biological sample, wherein the biological sample is of unknown or uncalibrated analyte

concentration.

**[0832]** In one preferred embodiment, a system for monitoring analyte concentration in a biological sample of a host is provided, the system including: a substantially continuous analyte sensor configured to produce a data signal indicative of an analyte concentration in a host during exposure of the sensor to a biological sample; a reference solution including a known analyte concentration, wherein the system is configured to expose the sensor to the reference solution, and wherein the system is configured to produce a data signal indicative of an analyte concentration in the reference solution during exposure of the sensor to the reference solution; and a computer system including programming configured to determine calibration information and calibrate a signal associated with a biological sample there from, wherein the calibration information is determined from a signal associated with exposure of the sensor to the reference solution and a signal associated with exposure of the sensor to a biological sample, wherein the biological sample is of unknown or uncalibrated analyte concentration.

**[0833]** In some embodiments, systems and methods are configured to process an impulse response of a signal associated with exposure of the sensor to a biological sample, wherein the biological sample is of unknown or uncalibrated analyte concentration, in order to determine an offset between a baseline measurement associated with a reference solution and a baseline measurement associated with a biological sample (e.g., $b_{offset}$).

**[0834]** **Fig. 11** is a graph that schematically illustrates a derivative of the step response shown in **Fig. 10. Fig. 11** can also be described, as the impulse response of the signal associated when a sensor is exposed to a step change to a biological sample of unknown or uncalibrated analyte concentration, in one exemplary embodiment. In this embodiment, the impulse response can be defined by a sum of two exponentials functions (e.g., $(ae^{-k1*t} - ae^{-k2*t})$, where k1 and k2 are time constants characteristic of the sensor), wherein the impulse response starts at 0 at t=0 and is expected to decay to 0 as t becomes large (as time passes). The impulse response reaches a peak, shown as point **1050** in **Fig. 11**, which represents the maximum rate of change of the associated signal (see **Fig. 10**, for example). Additionally, although it is expected that the signal will decay to 0 as t becomes large, **Fig. 11** illustrates a plateau **1052** above the y-axis; namely, wherein the plateau **1052** does not hit 0.

**[0835]** It has been discovered that the positive value **1054** of the plateau substantially represents the slope of the $b_{offset}$ artifact **1010** (**Fig. 10**). Accordingly, when the slope is drawn from 1-0 of the step response (see line **1012** of **Fig. 10**), the "y" value **1016** of that slope line at the end of the step response **1014**, represents $b_{offset}$. Accordingly, $b_{offset}$ can then be added to the equation y=mx+b (where $b= b_{reference} + b_{offset}$) and a conversion function (calibration factor) can be determined to calibrate the sensor system (i.e., using both steady state information and transient information and including using the signal associated with exposure of the sensor to a biological sample of unknown or uncalibrated analyte concentration.)

**[0836]** In some alternative embodiments, systems and methods are configured to process an impulse response (such as shown in **Fig. 11**) associated with a step change (such as shown in **Fig. 10**) to determine a time point of a steady state measurement during which an analyte concentration can be obtained. As described above, in some circumstances, it can be difficult to determine a steady state time point (e.g., **1006** in **Fig. 10**) at which time point the signal accurately represents the analyte concentration. Accordingly, systems and methods configured to determine the time point (e.g., **1006** in **Fig. 10**) in the step response associated with exposure of the sensor to a biological sample of unknown or uncalibrated analyte concentration have been discovered, which time point accurately represents the analyte concentration in the biological sample. Because the impulse response can by defined by exponentials (discussed above), systems and methods can be configured to process the exponential equation(s) with variable parameters to determine a best-fit to the impulse response curve determined from exposure of the sensor to the biological sample. It has been discovered that this best fit of the impulse response provides sufficient information to determine the time point **1056** (**Fig. 11**) at which the decay curve should have decayed to the y-intercept; namely, the time point **1056** where the decay curve should have hit y=0 indicates the (steady state) time point in the step response (e.g., **1006** in **Fig. 10**) that accurately represents the analyte concentration without the $b_{offset}$ artifact **1010**. Accordingly, (y=mx+b) can then be used to calibrate the sensor system, including the signal value "y" at the time indicated by the extrapolated impulse response curve (e.g., and using sensitivity and baseline information determined from one or more reference calibration solutions, such as described in more detail elsewhere herein.

**[0837]** In some other alternative embodiments, systems and methods are configured to compare steady state information and transient information for a plurality of time-spaced signals associated with biological samples of unknown or uncalibrated analyte concentration to determine an offset between a baseline measurement associated with a reference solution and a baseline measurement associated with the biological samples.

**[0838]** In some exemplary embodiments, $b_{offset}$ is determined by plotting level (i.e., the point at which the step response plateaus or ends) vs. rate (i.e., maximum rate of change of the step response determined from the peak of the impulse response curve) for a plurality of step responses (e.g., time-spaced signals) and drawing a regression line of the plotted points, such as described in more detail with reference to **Fig. 12**.

**[0839]** **Fig. 12** is a graph that illustrates level vs. rate for a plurality of time-spaced signals associated with exposure of the sensor to biological samples of unknown or uncalibrated analyte concentration. The y-axis represents maximum

rate of change for each step response; the x-axis represents level (signal level (e.g., in counts) obtained at the plateau of the signal and/or the end of the step response.) Each point **1080** on the plot represents level vs. rate for each of the plurality of time-spaced signals. A regression line **1082** is drawn using known regression methods, as is appreciated by one skilled in the art. The point **1084** at which the line **1082** crosses the y-axis represents the signal associated with a reference (e.g., 100 mg/dL calibration solution) plus $b_{offset}$ Accordingly, $b_{offset}$ can be determined by subtracting the signal associated with the reference from the point **1084** at which the line 1082 crosses the y-axis. Thus, $b_{offset}$ determined from the plot as described above, can be included in the equation y=mx+b (where $b= b_{reference} + b_{offset}$) and a conversion function (calibration factor) can be determined to calibrate the sensor system (i.e., using both steady state information and transient information and including using the signal associated with exposure of the sensor to a biological sample of unknown or uncalibrated analyte concentration.)

**[0840]** In some embodiments, $b_{offset}$ is an adjustable parameter, wherein the sensor system includes systems and methods configured to determine $b_{offset}$ with each measurement cycle (each time the sensor is exposed to the biological sample) and to adjust the calibration factor (conversion function), including $b_{offset}$ with each measurement cycle, responsive to a change in $b_{offset}$ above a predetermined threshold, and/or responsive to external information, for example.

**[0841]** In some embodiments, systems and methods are provided to detect a shift in the baseline and/or sensitivity of the signal based on a comparison of steady state information and transient information, such as described in more detail with reference to **Fig. 12.** In some embodiments, systems and methods are provided to correct for a shift in the baseline and/or sensitivity of the signal based on a comparison of steady state information and transient information. In some embodiments, systems and methods are provided to initiate a calibration responsive to detection of a shift in the baseline and/or sensitivity of the signal based on a comparison of steady state information and transient information.

**[0842]** Referring again to **Fig. 12**, regression line **1082** is shown for a selected plurality of time spaced signals. In some embodiments, multiple regression lines can be drawn for a plurality of different windows of time spaced signals (e.g., time-shifted windows). In these embodiments, a comparison of a regression line from a first window of time spaced signals as compared to a regression line drawn from a second window of time spaced signals can be used to diagnose a shift and/or drift in sensor sensitivity and/or baseline. For example, in **Fig. 12**, line **1082** represents a regression line drawn for a first window of data over a first period of time; dashed line **1086** represents a regression line drawn for a second window of data over a second period of time; and dashed line **1088** represents a regression line drawn for a third window of data over a third period of time. In this example, dashed line **1086** is shifted along the y-axis from the first line **1082,** indicating a drift or shift in the sensor's baseline from the first time period to the second time period; dashed line **1088** is shifted along the x-axis from the first line **1082,** indicating a drift or shift in the sensor's sensitivity from the first time period to the third time period. Accordingly, a shift in the regression line can be used to diagnose a shift or drift in the sensor's signal and can be used to trigger a corrective action, such as update calibration and/or re-calibration using any of the methods described herein. Additionally or alternatively, the shift in the line can be used to correct a shift or drift in the sensor's signal; for example, the amount of shift in the line can be used to update calibration accordingly (e.g., the change in y-value between two regression lines can be representative of a corresponding change in baseline between two time periods, and the calibration information updated accordingly). One skilled in the art appreciates that some combination of shift or drift of the baseline and sensitivity can occur in some situations, which can be similarly detected and/or corrected for.

Diagnostics and Fail-safes

**[0843]** In some embodiments, the system includes programming configured to diagnose a condition of at least one of the sensor and the host responsive to calibration information. In some embodiments, the system intermittently or continuously determines at least some calibration information (e.g., sensitivity information, $b_{offset}$, and the like), each time the sensor is exposed to a reference solution and/or a biological sample.

**[0844]** In one embodiment, systems and methods are configured to find a plateau and/or stable window of data in response to exposure of the sensor to at least one of a reference solution and a biological sample. In some embodiments, if the system cannot find the plateau and/or stable window of data, the system is configured to "fail-safe;" for example, in some circumstances, a lack of plateau and/or stable window of data may be indicative of dilution and/or mixture of the reference solution (e.g., calibration solution) with the biological sample (e.g., blood), and/or interruption/disruption of expected/desired fluid flow. Additionally, in some circumstances, a lack of plateau and/or stable window of data may be indicative of interfering species in the signal.

**[0845]** In general, the term "fail-safe" includes modifying the system processing and/or display of data in some manner responsive to a detected error, or unexpected condition, and thereby avoids reporting and/or processing of potentially inaccurate or clinically irrelevant analyte values.

**[0846]** In another embodiment, systems and methods are configured to process a signal responsive to exposure of the signal to a reference and/or biological sample to determine whether the signal is within a predetermined range; if the signal falls outside the range, the system is configured to fail-safe.

**[0847]** In some embodiments, systems and methods are configured to determine calibration information including sensitivity information, wherein the system includes programming configured to diagnose an error responsive to a change in sensitivity above a predetermined amount. For example, in a sensor system as described in more detail with reference to the exemplary embodiment of **Figs. 8A** to **8C**, the system can be configured to determine a sensitivity value during each calibration phase; and wherein the system can be configured to fail-safe when the sensitivity of a calibration phase differs from the previously stored sensitivity by more than a predetermined threshold. In this exemplary embodiment, fail-safe can include not using the sensitivity information to update calibration, for example. While not wishing to be bound by theory, the predetermined threshold described above allows for drift in the sensitivity of the sensor, but prevents large fluctuations in the sensitivity values, which may be caused by noise and/or other errors in the system.

**[0848]** In some embodiments, systems and methods are configured to diagnose error in the sensor system by ensuring the sensor signal (e.g., raw signal of the reference solution(s)) is within a predetermined range. In some embodiments, the sensor signal must be within a predetermined range of raw values (e.g., counts, current, etc). In some embodiments, one or more boundary lines can be set for a regression line drawn from the calibration phase. For example, subsequent to performing regression, the resulting slope and/or baseline are tested to determine whether they fall within a predetermined acceptable threshold (boundaries). These predetermined acceptable boundaries can be obtained from *in vivo* or *in vitro* tests (*e.g.*, by a retrospective analysis of sensor sensitivities and/or baselines collected from a set of sensors/patients, assuming that the set is representative of future data). U.S. Patent Publication No. US-2007-0197889-A1 describes systems and methods for drawing boundaries lines. In some embodiments, different boundaries can be set for different reference solutions.

**[0849]** In some embodiments, systems and methods are configured for performing diagnostics of the sensor system (e.g., continuously or intermittently) during exposure of the sensor to a biological sample, also referred to as the measurement phase, for example, such as described in more detail above with reference to **Figs. 8A** to **8C.** In some embodiments, diagnostics includes determination and/or analysis of $b_{offset}$. In some embodiments, systems and methods are provided for comparing sequential $b_{offset}$ values for sequential measurement phases. In some embodiments, the system includes programming configured to diagnose an error and fail-safe responsive to a change in the $b_{offset}$ above a predetermined amount. In some embodiments, the system includes programming configured to re-calibrate the sensor responsive to changes in the $b_{offset}$ above a predetermined amount. In some embodiments, the system includes programming configured to detect an interfering species responsive to a change in the $b_{offset}$ above a predetermined amount.

**[0850]** In some embodiments, the system includes programming configured to diagnose a condition of the host's metabolic processes responsive to a change in $b_{offset}$ above a predetermined amount. In some embodiments, the system includes programming configured to display or transmit a message associated with the host's condition responsive to diagnosing the condition. While not wishing to be bound by theory, it is believed that changes in $b_{offset}$ can be the result of an increase (or decrease) in metabolic by-products (electroactive species), which may be a result of wounding, inflammation, or even more serious complications in the host; accordingly, changes in $b_{offset}$ can be useful in diagnosing changes in the host's health condition.

**[0851]** In some embodiments, the system includes programming configured to detect sensor error, noise on the sensor signal, failure of the sensor, changes in baseline, and the like, responsive to a change in $b_{offset}$ above a predetermined amount.

**[0852]** In some embodiments, the system includes programming configured to determine a time constant of the sensor. One method for calculating a time constant for a sensor includes determining an impulse response to a step change, wherein time at the peak of the impulse response represents a time constant for the sensor. While not wishing to be bound by theory, it is believed that the time constant determined from the peak of the impulse response should remain substantially the same throughout the life of the sensor. However, if a shift in the time constant (between step changes and their associated impulse response curves) above a predetermined range is detected, it can be indicative of an unexpected sensor condition or error, for example. Accordingly, by comparing time constants from a plurality of impulse response curves (derived from a plurality of step responses), programming can be configured to diagnose a sensor condition or error and initiate programming (e.g., fail-safe), accordingly.

**[0853]** Accordingly, the system can "fail-safe," including performing one or more of the following fail-safe responses: temporarily or permanently suspending (e.g., discontinuing) display of analyte data, updating calibration or re-calibrating the sensor, requesting external reference values, using external reference value(s) as confirmation of a detected condition, using external reference value(s) to update calibration or re-calibrate the sensor, shutting the system down, processing the sensor data to compensate for the change in $b_{offset}$, transmitting one or more messages to the user interface or other external source regarding the sensor condition, and the like.

**[0854]** In preferred embodiments, the sensor electronics include a fail-safe module that is configured to detect a system malfunction. The fail-safe module can be configured to detect a variety of system malfunctions. For example, the fail-safe module can be configured to detect electrical malfunctions, malfunctions of the system fluidics, malfunctions of the sensor, and/or the like. For example, the system electronics can be configured to test and/or track the functions of different system components, and to intelligently recognize aberrant changed in such functions as possible malfunctions.

**[0855]** In some embodiments, the fail-safe module is configured to detect electrical malfunctions, such as but not limited to short circuit and electrical malfunction associated with start-up and/or sensor break-in. For example, it is known that an I/R drop across an electrochemical sensor's working electrode(s) generally occurs during use. The I/R drop can be monitored for changes, which can be indicative of an electrical malfunction. Similarly, during use of an analyte sensor in an intravascular system, wherein the sensor is periodically/alternately exposed to at least one reference solution and a blood sample, the generated signal is expected to rise and fall in a regular fashion, for example a recognizable waveform pattern including peaks and valleys. For example, when the enzyme sensor is exposed to blood, the rate of the step-response indicates the change in glucose from calibration solution to blood. The plateau of the response also indicates the glucose concentration (although the plateau generally indicates the absolute glucose concentration, not the change in glucose concentration). Accordingly, the fail-safe module can be configured to monitor the pattern of step-responses (e.g., over time) for substantial changes/deviations, which can be indicative of electrical malfunctions. For example, in some circumstances, the signal can "rail" or "zero," wherein the signal rapidly changes to a non-physiological analyte value and remains relatively stable at that non-physiological value for a period of time. In some circumstances, the rapid change in value occurs at a rate that is also not physiologically possible. In some circumstances, a comparison of the rate-response (e.g., transient/kinetic information) *versus* plateau response (e.g., steady state information) of each blood measurement can be used as an indicator of sensitivity drift. Similarly, in some embodiments, comparison of rate-response *versus* plateau response on a non-enzymatic electrode/sensor when expose to blood can provide information related to the sensor's response to non-glucose species in the sample, which can be used to determine if and/or when to fail-safe.

**[0856]** In some embodiments, the fail-safe module is configured to detect fluidics malfunctions, which can include a malfunction of any part of the system configured to contact and/or move a fluid. For example, the vascular access device (catheter) can become occluded, such as due to blood clotting or pressing the *in vivo* orifice (e.g., catheter tip) against a vessel wall such that fluid cannot move in an out of the vascular access device. In some circumstances, the tubing can become kinked, the host can move his arm such that the catheter becomes bent, bubbles may get into the system, and/or the flow control system can malfunction (e.g., not metering fluid flow into the host or not withdrawing blood samples as it was programmed to do), such that washing, calibration, sample collection and the like are not performed as programmed. For example, the sample can become diluted with calibration solution, there may be clotting on a portion of the sensor, and the like. In one exemplary embodiment, the fail-safe module is configured detect fluidics malfunctions by monitoring the pattern of signal increases/decreases generated on the working electrode(s), to detect periods of time during which the signal does not follow an expected waveform (e.g., using known waveform analysis methods and/or pattern recognition algorithms), such as when the signal hits upper and/or lower limits. For example, if the signal zeros for a period of time, the fluidics system may be unable to withdraw a blood sample due to kinking of the tubing or occlusion of the catheter. In another example, the signal on the working electrode(s) may stop going back down when the tested blood sample should be re-infused into the host and the sensor should be contacted with wash/reference solution, which is indicative of a malfunction of expelling the used sample, washing the sensor, and the like. For example, if the catheter becomes occluded by a blood clot, the flow control system will be unable to meter fluid through the catheter.

**[0857]** In some embodiments, the fail-safe module is also configured to detect malfunctions of the analyte sensor. For example, a sensor malfunction includes but is not limited to noise on the signal, drift of sensor sensitivity and/or baseline, a broken component of the sensor, blood clotting on a portion of the sensor, and cross-talk, as described in more detail elsewhere herein.

**[0858]** In preferred embodiments, the fail-safe module is further to evaluate a detected malfunction against a criterion. One or more criteria can be preprogrammed (including reprogrammed) into the system, such as be a host, a caretaker of the host, and/or the manufacturer. For example, in some embodiments, one or more criteria could be selected from a menu/list of criteria, or manually programmed into the system. Available criteria can be responsive to the different types of malfunctions can occur and/or a user-perceived level of severity and/or urgency.

**[0859]** After a system malfunction has been detected, the fail-safe module is configured to provide an alert, an alarm and /or an instruction, such as to the host and/or a caretaker of the host. For example, the system can be configured to provide an auditory alarm (e.g., beeps, busses, siren, pre-recorded voice message, etc.), a visual alert (e.g., blinking or flashing lights or display screen), a tactile alert (e.g., vibrating), transmitting alerts/instructions to a remote location, such as a nurse's station or a doctor's phone/PDA, and the like.

**[0860]** As a non-limiting example, a method is provided for detecting system malfunctions *via* the processing of the data of a continuous analyte sensor implanted in a host's circulatory system. As described elsewhere herein, the continuous analyte sensor is configured to generate a signal associated with an *in vivo* analyte concentration when the sensor is implanted in the host. Additionally, the sensor electronics comprises a fail-safe module configured to detect a system malfunction. Accordingly, the method includes exposing the sensor to a sample from the host's circulatory system and detecting a malfunction of the system, such as but not limited to an electrical malfunction, a fluidics malfunction, and/or a sensor malfunction. In some embodiments, an equilibrium analysis and/or a kinetic analysis of received data/signal is performed. As described elsewhere herein, the data include steady state (e.g., sensitivity and/or baseline

information) and/or transient state information, which can be evaluated/analyzed to provide alerts, alarms and/or instructions.

**[0861]** As described herein, the system (e.g., the processor module and/or fail-safe module) can be configured to analyze steady state and or transient state information generated during the step-up response of the sensor, such as during the switch from blood to reference solution and *vice versa,* also referred to as waveform analysis (for example, using pattern recognition algorithms, and the like). For example, by analyzing the step-up waveform of a dual-electrode sensors non-enzymatic working electrode, the quality of blood sample aspiration can be evaluated (e.g., fluidics). Similarly, by analyzing the step-down waveform of the sensor's non-enzymatic working electrode, the quality of washing can be evaluated (e.g., fluidics). As another example, analysis of direction of the step-up response of the plus-enzyme working electrode upon exposure to blood can used to evaluate calibration accuracy.

**[0862]** As a non-limiting example, wherein the system is configured to use a 100 mg/dl glucose reference solution and the plus-enzyme working electrode is exposed to blood having a glucose concentration greater than 100 mg/dL, the signal response should step up. When the plus-GOX working electrode is exposed to blood having a glucose concentration less than 100 mg/dL, the signal response should step down. If the system calculated a glucose value less than 100 mg/dL (e.g. 70 mg/dL) when the plus-GOX working electrode stepped up, a calibration error is detected and the system can fail-safe. Similarly, if the system calculated a glucose value greater than 100 mg/dL (140 mg/dL) when the plus-GOX working electrode stepped down, an error is detected and the system can fail-safe.

Signal Amplitude and/or Noise Amplitude Analysis and Fail-Safes

**[0863]** In some embodiments, the system is configured to fail-safe when analysis of signal amplitude and/or noise amplitude (residual) of the sensor's working electrode(s) indicates that the integrity of the system's electrical connections are compromised, such as at the sensor or at one or more components of the system's electronics. For example, in the case of a dual-electrode sensor, the correctness of the relationship of the plus-enzyme signal to the non-enzymatic signal can be evaluated by comparing signal amplitudes in calibration solution (or in blood). For example, the non-enzymatic signal being greater than the plus-enzyme signal can be indicative of a sensor malfunction, and the system can fail-safe. Simi

**[0864]** In some embodiments, sensor diagnostics evaluates whether the calculated sensitivity falls within a predetermined limit and/or range, for example, 1-10 count/mg/dL, as described in more detail elsewhere herein. In some embodiments, the sensor diagnostics evaluates a difference in signal (raw, filtered and/or calibrated) between the non-enzymatic working electrode and the enzymatic working electrode (e.g., in reference/calibration solution) to determine whether they fall within predetermined limits and/or ranges. As one example, a sensor/calibration diagnostic evaluates the blood baseline of the enzymatic and non-enzymatic electrodes to determine whether they fall within a predetermined limit/range.

**[0865]** In some embodiments, the system is configured to use *a priori* information to set tolerances for sensitivity and/or baseline drift. Preferably, the system is configured to discriminate between real/actual changes in sensitivity/baseline from signal artifacts, such as by using tolerance limits based on historical/empirical data. For example, in some embodiments, if the sensitivity and/or baseline drifts outside predetermined limits and the drift is not a transient artifact (e.g., it persists), an error can be detected in sensor function and/or calibration.

**[0866]** In yet another alternative embodiment of signal artifacts detection that utilizes examination or evaluation of the signal information content, filtered (e.g., smoothed) data is compared to raw data (e.g., in sensor electronics or in receiver electronics). In one such embodiment, a signal residual is calculated as the difference between the filtered data and the raw data. For example, at one time point (or one time period that is represented by a single raw value and single filtered value), the filtered data can be measured at 50,000 counts and the raw data can be measured at 55,500 counts, which would result in a signal residual of 5,500 counts. In some embodiments, a threshold can be set (e.g., 5000 counts) that represents a first level of noise (e.g., signal artifact) in the data signal, when the residual exceeds that level. Similarly, a second threshold can be set (e.g., 8,000 counts) that represents a second level of noise in the data signal. Additional thresholds and/or noise classifications can be defined as is appreciated by one skilled in the art. Consequently, signal filtering, processing, and/or displaying decisions can be executed based on these conditions (e.g., the predetermined levels of noise).

**[0867]** Although the above-described example illustrates one method for determining a level of noise, or signal artifact(s), based on a comparison of raw *vs*. filtered data for a time point (or single values representative of a time period), a variety of alternative methods are contemplated. In an alternative exemplary embodiment for determining noise, signal artifacts are evaluated for noise episodes lasting a certain period of time. For example, the processor (in the sensor or receiver) can be configured to look for a certain number of signal residuals above a predetermined threshold (representing noise time points or noisy time periods) for a predetermined period of time (e.g., a few minutes to a few hours or more).

**[0868]** In one exemplary embodiment, a processor is configured to determine a signal residual by subtracting the filtered signal from the raw signal for a predetermined time period. It is noted that the filtered signal can be filtered by

any known smoothing algorithm such as described herein, for example a 3-point moving average-type filter. It is further noted that the raw signal can include an average value, e.g., wherein the value is integrated over a predetermined time period (such as 5-minutes). Furthermore, it is noted that the predetermined time period can be a time point or representative data for a time period (e.g., 5 minutes). In some embodiments, wherein a noise episode for a predetermined time period is being evaluated, a differential can be obtained by comparing a signal residual with a previous signal residual (e.g., a residual at time (t)=0 as compared to a residual at (t)-5 minutes.) Similar to the thresholds described above with regard to the signal residual, one or more thresholds can be set for the differentials, whereby one or more differentials above one of the predetermined differential thresholds defines a particular noise level. It has been shown in certain circumstances that a differential measurement as compared to a residual measurement as described herein, amplifies noise and therefore may be more sensitive to noise episodes, without increasing false positives due to fast, but physiological, rates of change. Accordingly, a noise episode, or noise episode level, can be defined by one or more points (e.g., residuals or differentials) above a predetermined threshold, and in some embodiments, for a predetermined period of time. Similarly, a noise level determination can be reduced or altered when a different (e.g., reduced) number of points above the predetermined threshold are calculated in a predetermined period of time.

**[0869]** In some embodiments, the amplitude of total signal, which can also be described as power of the total signal, analyte signal (with or without baseline (e.g., non-constant noise)), and/or non-constant noise, is periodically or continuously obtained using methods such as are described in more detail elsewhere herein (e.g., RMS method), wherein the amplitude is a measure of the strength of the signal component. In some embodiments, signal artifact events are detected by analysis of amplitudes of various signal components, such as the amplitude of the non-constant noise component as compared to the amplitude of the analyte signal (with or without baseline).

**[0870]** In some embodiments, a start of a signal artifact event is determined when the amplitude (power) of a signal artifact meets a first predetermined condition. In one embodiment, the first predetermined condition includes a residual amplitude of at least about 2, 3, 4, 5, 6, 7, 8, 9, 10, 12, 14, 16, 18, 20 or 25% of the total signal or analyte signal amplitude (with or without baseline). In another embodiment, the first predetermined condition includes a differential amplitude (amplitude of a differential) of at least about 2, 3, 4, 5, 6, 7, 8, 9, 10, 12, 14, 16, 18, 20 or 25% of the total signal or analyte signal amplitude (with or without baseline). In some embodiments, the first predetermined condition includes a plurality of points (e.g., non-constant noise signal, residual, or differential) within a predetermined period (e.g., 5, 10, 30, or 60 minutes) above a predetermined threshold (e.g., an amplitude or a percentage amplitude), wherein the plurality of points includes 2, 3, 4, 5, 6, 7, 8 or more values.

**[0871]** In some embodiments, an end of a signal artifact event is determined when then the amplitude (power) of a signal artifact meets a second predetermined condition. In one embodiment, the second predetermined condition includes a residual amplitude of no more than about 2, 3, 4, 5, 6, 7, 8, 9, 10, 12, 14, 16, 18, 20 or 25% of the total signal or analyte signal amplitude (with or without baseline). In another embodiment, the second predetermined condition comprises a differential amplitude of no more than about 2, 3, 4, 5, 6, 7, 8, 9, 10, 12, 14, 16, 18, 20 or 25% of the total signal or analyte signal amplitude (with or without baseline). In some embodiments, the second predetermined condition includes a plurality of points (e.g., non-constant noise signal, residual, or differential) within a predetermined period (e.g., 5, 10, 30, or 60 minutes) below a predetermined threshold (e.g., an amplitude or a percentage amplitude), wherein the plurality of points includes 2, 3, 4, 5, 6, 7, 8 or more values. Additional description can be found in US Patent Publication Nos. US-2005-0043598-A1, US-2007-0032706-A1, US-2007-0016381-A1, and US-2007-0033254-A1.

Auto-Calibration

**[0872]** Some preferred embodiments are configured for auto-calibration of the sensor system, wherein "auto-calibration" includes the use of one or more internal references to calibrate the sensor system. In some embodiments, auto-calibration includes systems and methods configured to calibrate the sensor based solely on internal reference values. For example, the system can be configured to obtain internal reference values by exposing the sensor to one, two or more calibration/reference solution, as described elsewhere herein. As another example, the sensor can be configured with a reference sensor disposed adjacent to the continuous analyte sensor, wherein both the continuous analyte and reference sensors obtain internal reference values for a substantially constant reference analyte (e.g., $O_2$), which are then used to calibrate the continuous analyte sensor (see the sections entitled "Optical Detection" and "Calibration Systems and Methods of Calibrating Dual-Electrode Sensors" above). However, in some alternative embodiments of auto-calibration, one or more external reference values can be used to complement and/or confirm calibration of the sensor system.

**[0873]** In some preferred embodiments, the system is configured to intermittently expose the sensor to a biological sample; however configurations of the sensor system that allow continuous exposure of the sensor to the biological sample are contemplated. In some embodiments, the system is configured to intermittently or periodically expose the sensor to a reference, however configurations of the sensor system that allow one or more independent or non-regular reference measurements initiated by the sensor system and/or a user are contemplated. In the exemplary embodiment

of a sensor system such as described with reference to **Figs. 8A** to **8C**, the system is configured to cycle between a measurement phase and calibration phase (with optional other phases interlaced therein (e.g., flush and KVO)).

**[0874]** In general, timing of auto-calibration can be driven by a variety of parameters: preset intervals (e.g., clock driven) and/or triggered by events (such as detection of a biological sample at a sensor). In some embodiments, one or more of the phases are purely clock driven, for example by a system configured to control the timing housed within a flow control device, remote analyzer, and/or other computer system. In some embodiments, one or more of the phases are driven by one or more events, including: exposure of the sensor to a biological sample (e.g., blood) exposure of sensor to a reference (e.g., calibration solution), completion of calibration measurement, completion of analyte measurement, stability of signal measurement, sensor for detecting a biological sample, and the like.

**[0875]** In one exemplary embodiment, calibration and measurement phases are driven by cleaning of sensor; namely, systems and methods are configured to detect when the sensor is in the biological sample and/or in the reference solution (e.g., calibration solution), wherein the system is configured to switch to appropriate phase responsive to detection of that sample/solution.

**[0876]** In another exemplary embodiment, an AC signal is placed on top of a DC signal (e.g., in an amperometric electrochemical analyte sensor), wherein systems and methods are configured to analyze an impedance response to the AC signal and detect a biological sample thereby.

**[0877]** In yet another exemplary embodiment, systems and methods are configured for analyzing the sensor's signal, wherein a change from a known reference solution (e.g., a known analyte concentration) can be detected on the signal, and the switch from the calibration phase to the measurement phase occur responsive thereto; similarly, the system can be configured to switch back to the calibration phase responsive to detection of the known signal value associated with the reference solution.

**[0878]** In yet another exemplary embodiment, systems and methods are configured to switch between phases responsive to one or more sensors configured to detect the biological sample and/or reference solution at a particular location.

**[0879]** In some embodiments, the sensor system is partially or fully controlled by a clock (e.g., predetermined time intervals), which timing can be confirmed by any of the events (e.g., triggers or sensors) described above.

**[0880]** In one exemplary embodiment, systems and methods are provided to enable auto-calibration of an integrated glucose sensor system with minimal user interaction. In this exemplary embodiment, the integrated sensor system is provided with the components described above, including a fluids bag, a flow control device, IV tubing, a flow control device, a remote analyzer, a local analyzer and a sensor/catheter, for example. At system start-up, a health care worker inserts the catheter and sensor into a host and injects a first reference solution (e.g., zero glucose saline solution) into the IV tubing, wherein the system is configured to allow a predetermined time period (e.g., 20 minutes) for the first reference solution to pass through the IV tubing and into the catheter. Subsequently, the health care worker couples the fluids bag to the IV tubing, wherein the fluids bag includes a second reference solution (e.g., 100 mg/dl glucose solution) configured to follow the first reference solution in the IV line. After injecting the first reference solution and coupling the second reference solution, the health care worker initiates the integrated sensor system (e.g., through the remote analyzer touch screen) after which the integrated sensor system automatically calibrates and functions for 24 hours without necessary user interface (for system calibration and/or initiation). In some embodiments, the sensor system is re-calibrated every 24 hours by injection of a new first reference solution (e.g., zero glucose saline solution).

**[0881]** In the above-described exemplary embodiment, the system is configured to calibrate the sensor with the first and second reference solution and using the methods described in the section entitled, "Systems and Methods for Processing Sensor Data." Additionally, the system is configured to automatically detect the difference in signal associated with the first and second reference solutions, for example, through steady state detection of a difference in signal level.

Examples

Example 1: Glucose Sensor System Trial in Dogs

**[0882]** Referring now to **Fig. 4**, glucose sensor systems of the embodiment shown in Fig. 1 were tested in dogs. The glucose sensors were built according to the preferred embodiments described herein. Namely, a first sensor (Test 1) was built by providing a platinum wire, vapor-depositing the platinum with Parylene to form an insulating coating, helically winding a silver wire around the insulated platinum wire (to form a "twisted pair"), masking sections of the electroactive surface of the silver wire, vapor-depositing Parylene on the twisted pair, chloridizing the silver electrode to form a silver chloride reference electrode, and removing a radial window on the insulated platinum wire to expose a circumferential electroactive working electrode surface area thereon, this assembly also referred to as a "parylene-coated twisted pair assembly."

**[0883]** An electrode domain was formed over the electroactive surface areas of the working and reference electrodes by dip coating the assembly in an electrode solution and drying. An enzyme domain was formed over the electrode

domain by subsequently dip coating the assembly in an enzyme solution and drying. A resistance domain was formed over the enzyme domain by spraying the resistance domain solution on the sensor construct.

**[0884]** After the sensor was constructed, it was placed in the protective sheath and then threaded through and attached to the fluid coupler.

**[0885]** A second sensor (Test 2) was constructed in the same manner as the first, except that the silver wire was disposed within (e.g., coiled within) the fluid coupler. Accordingly, only the platinum working electrode (a single wire) was inserted into the catheter during the experiment.

**[0886]** Prior to use, the sensors were sterilized using electron beam.

**[0887]** The forelimb of an anesthetized dog (2 years old, -40 pounds) was cut down to the femoral artery and vein. An arterio-venous shunt was placed from the femoral artery to the femoral vein using 14 gauge catheters and 1/8-inch IV tubing. A pressurized arterial fluid line was connected to the sensor systems at all times. The test sensor systems (test 1 and test 2) included a 20 gauge x 1.25-inch catheter and took measurements every 30 seconds. The catheter was aseptically inserted into the shunt, followed by insertion of the sensor into the catheter. A transcutaneous glucose sensor (control) of the type disclosed in U.S. Publ. No. US-2006-0155180-A1 was built and placed in the dog's abdomen according to recommended procedures. The dog was challenged with continuous incremental IV infusion of a 10% dextrose solution ("glucose challenge") until the blood glucose concentration reached about 400 mg/dL.

**[0888]** **Fig. 4** shows the experimental results. The thick line represents data collected from the Test 1 sensor. The thin line represents data collected from the Test 2 sensor. Diamonds represent data collected from a hand-held blood glucose meter (SMBG) sampled from the dog's abdomen. Raw glucose test data (counts) are shown on the left-hand Y-axle, glucose concentrations for the "SMBG" controls are shown on the right-hand y-axle, and time is shown on the X-aacle. Each time interval on the X-axle represents 29-minutes (e.g., 10:04 to 10:33 equals 29 minutes). Immediately upon insertion into a catheter, each test sensor began collecting data with substantially no sensor equilibration time (e.g., break-in time). Each test sensor responded to the glucose challenge substantially similarly to the control sensor. For example, each device shows the glucose signal increasing from about 3200 counts at 10:33 to about 6000-6700 counts at 11:31. Then, each device showed a rapid decrease in glucose signal, to about 4700 counts at 12:00. Additionally, the response of the test sensors and the control sensor were substantially similar (e.g., the majority of the test data was substantially equivalent to the SMBG data at each time point). From these experimental show that an indwelling glucose sensor system (as described herein) in contact with the circulatory system can provide substantially continuous glucose monitoring in a clinical setting.

Example 2: Glucose Sensor System Trial in Pigs

**[0889]** Referring now to **Fig. 5**, four glucose sensor systems of the embodiment shown in **Fig. 1** were tested in a pig (~104 lb), using the protocol described for Example 1, above. Glucose was continuously infused at increasing rates through a distally placed IV catheter until a readout of 300-400 mg/dl blood glucose was achieved (total 300 ml of a 10% dextrose IV solution). **Fig**. **5** shows the experimental results. Lines indicated the data from the four sensors (Test 1 through Test 4). Diamonds represent control measurements made with a hand-held glucose meter (SMBG). Raw glucose test data (counts) are shown on the left-hand Y-axle, glucose concentrations for the "SMBG" controls are shown on the right-hand y-axle, and time is shown on the X-axle. Test results show that though the sensors varied in sensitivity, each test sensor responded to glucose challenge substantially similarly to the control sensor (SMBG). These experimental results show that an indwelling glucose sensor system (of the preferred embodiments) in contact with the circulatory system can substantially continuously track glucose in a clinical setting.

Example 3: Glucose Sensor System with Flow Control Device Trial in Pigs

**[0890]** Referring now to **Fig. 13**, a glucose sensor was built according to the preferred embodiments described herein. Namely, a test sensor was built by providing a platinum wire, vapor-depositing the platinum with Parylene to form an insulating coating, helically winding a silver wire around the insulated platinum wire (to form a "twisted pair"), masking sections of the electroactive surface of the silver wire, vapor-depositing Parylene on the twisted pair, chloridizing the silver electrode to form a silver chloride reference electrode, and removing a radial window on the insulated platinum wire to expose a circumferential electroactive working electrode surface area thereon, this assembly also referred to as a "parylene-coated twisted pair assembly."

**[0891]** An electrode domain was formed over the electroactive surface areas of the working and reference electrodes by dip coating the assembly in an electrode solution and drying. An interference domain was formed over the electrode domain by subsequently dip coating the assembly in an interference domain solution and drying. An enzyme domain was formed over the interference domain by subsequently dip coating the assembly in an enzyme solution and drying. A resistance domain was formed over the enzyme domain by spraying the resistance domain solution on the sensor construct.

**[0892]** The test sensor was then located within a 20 gauge catheter and inserted in the femoral vein of a non-diabetic pig. The catheter was connected to an integrated sensor system **600** of the preferred embodiments. The flow control device **604** (e.g., a roller valve as depicted in **Figs. 8A-8C**) was configured to move between steps one and two, as described in the section entitled "Flow Control Device Function," above. A 107-mg/dL glucose solution was used to calibrate the sensors (e.g., flows from the reservoir **602**, through the tubing **606**, to the catheter 12). To mimic a diabetic's hyperglycemic state, a gradual infusion of 26% dextrose was given, until the pig's blood glucose was about 600 mg/dl. Then to mimic a hypoglycemic state, 10 U Humulin N was given, until the pig's blood glucose was about 50 mg/dl. Then, the pig's blood glucose was raised to about 100 mg/dl by a second 26% dextrose infusion.

**[0893]** **Fig. 13** is a graphical representation showing uncalibrated glucose sensor data and corresponding blood glucose values over time in a pig. Raw counts are represented on the left Y-axis. Glucose concentration is shown on the right Y-axis. Time is shown on the X-axis. Test measurements (e.g., measurements of blood glucose concentration obtained with the test sensor, raw counts) are shown as small, black dots. Control measurements (e.g., jugular vein blood samples analyzed on a Yellow Springs Instrument (YSI) glucose analyzer) are shown as diamonds.

**[0894]** During the experiment, the system was configured to alternate between calibration measurements (with the 107 mg/dl glucose solution) and blood glucose measurements, as described in the sections entitled "Step one: Contacting Sensor with Calibration Solution and Calibration" and "Step Two: Sample Collection and Measurement," respectively. Accordingly, as the experiment proceeded the test signal oscillated between calibration solution (107 mg/dl) and blood glucose measurements. The sensor (test) blood glucose measurement correlated tightly with the control blood glucose measurements. For example, as the pig's blood glucose concentration increased (due to infusion of glucose), so did the test measurements, reaching about 550 mg/dl at about 12:20. Similarly, as the pig's blood glucose concentration decreased (due to infusion of insulin), so did the test measurements, decreasing to about 50 mg/dl at about 14:45.

**[0895]** From these data, it was concluded that a glucose sensor system of the preferred embodiments (including a valve as described with reference to **Figs. 8A** to **8C**) accurately and sensitively measures intravenous glucose concentration over a wide dynamic range.

Example 4: Glucose Sensor System with Flow Control Device Trial in Humans

**[0896]** Referring now to **Fig. 14**, a glucose sensor, constructed as described in Example 3, and an integrated sensor system (as described in Example 3) were tested in a volunteer, diabetic host. The flow control device was configured as shown in **Figs. 8A-8C.** The system was configured to alternate between a calibration phase and a blood glucose measurement phase, as described elsewhere herein. At sensor/catheter initialization, a 0 mg/dl glucose saline solution filled syringe was injected into the IV tubing and the fluids bag including 100 mg/dl glucose heparinized saline solution was subsequently coupled to the tubing. The system was then turned on (e.g., sensor initialized). The 0 mg/dl glucose saline solution passed over the sensor, after which the 100 mg/dl glucose heparinized saline solution subsequently passed over the sensor allowing for initial calibration information to be collected. The system, including a flow control device as described with reference to **Figs. 8A** to **8C**, then oscillated between exposure of the sensor to a blood sample and exposure of the sensor to the 100 mg/dl glucose heparinized saline solution. The sensor auto-calibrated by a combination of calibration information obtained from measurement of the 0 mg/dl-glucose and 100 mg/dl-glucose saline solutions and the step-change-response of the sensor to the blood sample, according to the methods described in the section entitled "Systems and methods for Processing Sensor Data." No external measurements (e.g., blood glucose measurements by YSI or finger stick) were used to calibrate the system in this example. During the experiment, the flow control device cycled between step one (measuring the 100 mg/dl-glucose solution) and step two (blood sample take up and measuring the blood glucose concentration), such that one cycle was completed every 5-minutes. The experiment was conducted for a period of about 2.5 days. The host followed her usual schedule of meals and insulin injections.

**[0897]** **Fig. 14** is a graphical representation showing calibrated venous blood glucose sensor measurements (test, black dots) and corresponding control blood glucose measurements (YSI, large circles) over time in the volunteer diabetic host. Glucose concentration is shown on the Y-axis and time on the X-axis. Test measurements tracked closely with control measurements, ranging from about 350 mg/dl, at about 10:00 and about 15:30, to about 50 mg/dl, at about 11:45. From these data, it has been concluded that 1) the sensor calibration methods of the preferred embodiments accurately calibrate the sensor and 2) the glucose sensor system of the preferred embodiments accurately measures intravenous glucose concentration over a wide dynamic range, for two or more days, in humans.

Example 5: Detection of Crosstalk *In Vitro*

**[0898]** In general, crosstalk in dual-electrode sensors can be examined by recording the signal detected at each working electrode while placing them in a series of analyte solutions. This example describes one exemplary *in vitro* test for crosstalk on a dual-electrode analyte sensor. A variety of dual-electrode analyte sensors can be tested for crosstalk, using this method.

**[0899]** First, the sensor to be tested is placed in a phosphate buffered saline (PBS) for several minutes, such as until a stable signal is detected from both working electrodes. Next, the sensor is challenged with glucose solutions and optionally with one or more known noise-causing substances. For example, sensor can be placed sequentially in a series of glucose solutions (e.g., 40-mg/dl, 200-mg/dl and 400-mg/dl glucose in PBS) and the signals from the two working electrodes graphed.

**[0900]** If there is no crosstalk between the working electrodes, then, as the sensor is placed in solutions of increasingly higher glucose concentration, the graphed signal of the Plus GOx electrode should show corresponding signal increases, while the No GOx electrode signal should exhibit little or no change in signal. However, when the sensor is placed in a solution of a known noise-causing species (e.g., acetaminophen, ibuprofen, vitamin C, urea, and the like), both working electrodes (Plus GOx and No GOx) should exhibit an increase in signal. In some circumstances, this increase in signal is a dramatic spike in signal.

**[0901]** If there is crosstalk, then the signals from both electrodes should increase as the sensor is moved to solutions of increased glucose concentration. Similarly, when the sensor is placed in a solution of a known noise-causing species, both working electrodes should exhibit an increase in signal.

Example 6: Effect of Electroactive Surface Size in Dual-Electrodes *In Vivo*

**[0902]** The effect of size of the electroactive surfaces (of the working electrodes) on noise measurement was examined in non-diabetic human hosts. Sensors having electroactive surfaces of different sizes, and lacking GOx in the enzyme layer were constructed as follows. Clean, insulated Pt/Ir wires were separated into two groups. For **Group 1,** 0.029" of the insulation was removed from each wire (e.g., about its entire circumference), to expose electroactive surfaces. For **Group 2,** the same procedure was performed, except that two sequential 0.029" portions of the insulation were removed; effectively doubling the size of the **Group 2** electroactive surfaces relative to those of **Group 1.** After exposure of the electroactive surfaces, the two groups of wires were treated identically. On each wire, a portion of the sensor's membrane was fabricated by the sequential application (and curing thereof) of electrode domain, enzyme domain and resistance domain materials. The enzyme domain material contained no active GOx, so that the sensors would be able to detect only noise (no analyte). Next, pairs of wires (e.g., two **Group 1** wires or two **Group 2** wires) were aligned such that the electroactive surfaces were parallel to each other, and then twisted together. An Ag/AgCl ribbon was wrapped around a portion of the twisted wires (to form the reference electrode), and then additional resistance domain material was applied to the assembly. Each host consumed 1,000-mg of acetaminophen near the end of the trial, so that the affect of a known interferent could be examined.

**[0903]** **Fig. 15A** is a graph illustrating the *in vivo* experimental results from implantation of a **Group 1** sensor (one 0.029" electroactive surface area per electrode). The Y-axis represents raw counts and the X-axis represents time. The data collected from each electroactive surface is shown as a line (**E1, No GOx** and **E2, No GOx**). Please note that the data represents only the noise component of the signal. No analyte component was detected, due to the lack of GOx in the portions of the membrane adjacent to the electroactive surfaces. Referring now to **Fig. 19A**, the noise signal detected by each of the working electrodes (**E1, E2**) fluctuated widely throughout the implantation time. The amplitude of the noise component detected by **E2** was substantially greater than the noise component detected by **E1.** For example, referring to the data circled by oval **A**, the noise signal peaks detected by **E2** were substantially greater than those detected by **E1.** Additionally, the noise fluctuations between the two working electrodes were not always in the same direction. For example, referring to the data circled by oval **B**, the **E2** noise signal component increased while the **E1** noise signal component decreased. When the sensor was challenged with acetaminophen (drug consumption indicated by the arrow), both electrodes registered a substantial increase in noise signal, wherein the shapes of the curves were substantially similar. However, **E1** detected a substantially lower total amount of noise when compared with that detected by **E2**; this difference in signal amplitude (in response to a non-biologic interferent) indicates that the signals (e.g., on the two working electrodes) did not have substantially similar sensitivities.

**[0904]** **Fig. 15B** is a graph illustrating the *in vivo* experimental results from implantation of a **Group 2** sensor (two 0.029" electroactive surface areas per electrode). As described above, the **Group 2** electroactive surfaces are about two-times as large as those of the **Group 1** sensors. As before, the Y-axis represents raw counts and the X-axis represents time, and the data collected from each electroactive surface is shown as a line (**E1, No GOx** and **E2, No GOx**). As before, the data represents only the noise component of the signal. No analyte component was detected, due to the lack of GOx in the portions of the membrane adjacent to the electroactive surfaces. Referring now to **Fig. 15B**, the noise signal detected by each of the working electrodes (**E1**, **E2**). Throughout the course of the experiment (-20-hours), noise signals detected by **E1** and **E2** tracked very closely throughout the entire experiment. Namely, the noise signals detected by **E1** and **E2** were of substantially the same amplitude and followed substantially similar fluctuations, varying from about 7,000 counts (at 4:00 PM) to about 4500 counts (from about 6:30 AM to about 9:00 AM). Even when the **Group 2** sensor was challenged with acetaminophen (a known non-biological interferent that should cause a false signal on the sensor), the working electrodes recorded substantially equal signal amplitudes (-6,900 counts) and followed substantially similar

wave forms having substantially equivalent amplitudes; these data indicate that the two working electrodes had substantially equal sensitivities.

**[0905]** From these data, the inventors concluded that the electroactive surfaces of a dual-electrode glucose sensor must be sufficiently large in order for the two electrodes to detect substantially equal noise signal components. For example, in this experiment, the electroactive surfaces of the **Group 1** working electrodes, which did not measure noise equivalently, were 0.029" (e.g., along the electrode's length); while electroactive surfaces of the **Group 2** working electrodes, which did measure noise substantially equivalently, were two times as large (e.g., 2x0.029"=0.058" along the electrode's length) as those of the Group 1 working electrodes.

Example 7: Effect of Electroactive Surface Pacing in Dual-electrodes *In Trivo*

**[0906]** The effect of the spacing of the electroactive surfaces (of the working electrodes) on noise measurement was examined in non-diabetic human hosts. Sensors having two different configurations were built and tested. Sensors of **Configuration 1** (**Config. 1**) included Plus GOx and No GOx working electrodes with non-aligned (e.g., miss-aligned, skewed) electroactive surfaces. In other words, the electroactive surfaces were spaced such that, in the completed sensor, one electroactive surface would be more proximal to the sensor's tip than then other. Sensors of **Configuration 2** (**Config. 2**) also included Plus GOx and No GOx working electrodes, except that the electroactive surfaces were closely aligned (e.g., parallel). In **Config. 2**, the membrane was the insulator between the two working electrodes, enabling the very close spacing (i.e., the thickness of the membrane determined the spacing between the two working electrodes, between about 0.001 inches to about 0.003 inches between the two working electrodes.)

**[0907]** **Config. 1** sensors were fabricated as follow. For each sensor, two clean, insulated Pt/Ir wires were wound together (and an Ag/AgCl ribbon twisted there around), followed by removal of a portion of the insulating material from each wire to create the electroactive surfaces. The electroactive surfaces were offset (e.g., not next to each other) relative to the sensor's tip. The twisted wire pairs were then dipped in enzyme domain solution (including GOx) just far enough such that only the electroactive surface closest to the tip of the sensor was coated with the enzyme domain material (e.g., **E1, Plus GOx**). The electroactive surface farthest from the sensor tip was not coated with the enzyme domain material (e.g., **E2, No GOx**). After curing, resistance domain material was applied to the twisted pairs of wires.

**[0908]** **Config. 2** sensors were fabricated as follow. Clean, insulated Pair wires were divided into two groups. Electrode and enzyme (Plus GOx) domain materials were sequentially applied to the **E1, Plus GOx** working electrode wires. Electrode and enzyme (No GOx) domain materials were sequentially applied to the **E2, No GOx** working electrode wires. Resistance domain material was applied to all wires individually (e.g., to form independent/discontinuous first and second resistance domains). After the resistance domain material was cured, one each of the **E1, Plus GOx** and **E2,** No **GOx** were placed together such that the wires' electroactive surfaces were aligned, and then twisted together to form a twisted pair. An Ag/AgCl ribbon was wrapped around each twisted pair (but not covering the electroactive surfaces), followed by application of a continuous resistance domain (e.g., a third resistance domain) over the sensor. The resulting sensors included a coaxial helix configuration similar to that shown in **Fig. 18** of co-ending US Patent Application 11/865,572, filed October 1, 2007 and entitled "DUAL ELECTRODE SYSTEM FOR A CONTINUOUS ANALYTE SENSOR".

**[0909]** **Fig. 16A** is a graph illustrating the *in vivo* experimental results from implantation of a **Config. 1** sensor (non-aligned electroactive surfaces). The Y-axis represents raw counts and the X-axis represents time. The **E1, Plus GOx** electrode detected both glucose and noise signal components while the **E2, No GOx** electrode detected only the noise signal component. Throughout most of the experiment's duration, the two working electrodes recorded signals having somewhat similar waveforms with the two lines being relatively flat with little fluctuation in amplitude (the volunteer was not diabetic, and thus would generally not have large fluctuations in glucose level). During the acetaminophen challenge, the **E1, Plus GOx** signal rapidly peaked at about 12,000 counts and then gradually declined, while the **E2, No GOx** signal peak was much lower in amplitude (~6,000 counts).

**[0910]** **Fig. 16B** is a graph illustrating the *in vivo* experimental results from implantation of a **Config. 2** sensor (aligned electroactive surfaces). The Y-axis represents raw counts and the X-axis represents time. In general, the **E1, Plus GOx** electrode detected both glucose and noise signal components while the **E2, No GOx** electrode detected only the noise signal component. Throughout most of the experiment's duration, the two electrodes recorded signals having substantially similar waveforms; though the **E1, Plus GOx** electrode signal was generally higher in amplitude than that of the **E2, No GOx** electrode. When the sensor was challenged with acetaminophen, the signals of both working electrodes rapidly peaked at about 11,000-12,000 counts (e.g., the amplitudes of the two peaks were substantially equivalent/similar) and then gradually declined.

**[0911]** From these data, the inventors concluded that the electroactive surfaces of a dual-electrode glucose sensor must be sufficiently close together in order for the two electrodes to detect substantially equivalent noise signal components. Additionally, the inventors concluded that for a dual-electrode sensor including the combination of a continuous resistance domain disposed over discontinuous resistance domains (e.g., applied independently to the two working electrodes) the detected signal amplitudes more closely correspond to each other. This improves mathematical noise

correction by enabling better noise signal subtraction.

Example 8: Use of Dual-Electrode Glucose Sensors in Intravenous Porcine Model

**[0912]** Six dual-electrode glucose sensor IV systems, including a dual-electrode glucose sensor (e.g., comprising a first working electrode **E1** (e.g., configured to generate signals associated with both glucose and non-glucose-related species having oxidation/reduction potentials that overlap with that of glucose) and a second working electrode E2 (e.g., configured to generate signals associated with non-glucose-related species having oxidation/reduction potentials that overlap with that of glucose)) and a fluid control system (including a catheter inserted into a peripheral vein) were tested in a porcine model, as described elsewhere herein. A 100-mg/dl-glucose solution was used to calibrate and wash the dual-electrode sensor, according to procedures described elsewhere herein. To obtain an increase in serum glucose to hyperglycemic ranges, a 22.5% dextrose solution was gradually infused through the animal's left ear vein. The infusion was stopped once the blood glucose levels reached sufficiently high blood glucose values and a return to normal values was followed. Blood samples were collected in 5-minute intervals from the right jugular vein throughout the study. Plasma glucose values were determined on an YSI instrument. The study took about 4-hours.

**[0913]** Fig. 17A is a graph illustrating test results from an exemplary dual-electrode continuous glucose sensor coupled to a flow control system *via* a catheter inserted in the animal's femoral vein. The Y-axis represents glucose values (mg/dl) and the X-axis represents time. The small triangle represents glucose values provided by the dual-electrode glucose sensor system every 5-minutes, while the small diamond represents substantially time-corresponding control glucose values obtained from the YSI device.

**[0914]** **Fig. 17B** is a Clark Error Grid quantifying the clinical accuracy of the blood glucose estimates generated by the dual-electrode glucose sensor system as compared to a reference value (e.g., YSI). All values fall within area "A," which indicates that all values provided by the dual-electrode sensor are within 20% of the values provided by the YSI control test device. $R^2 = 0.9799$.

**[0915]** The dual-electrode glucose sensor tracked the animal's glucose values, from about 50-mg/dl to about 300-mg/dl over a period of about 4-hours. The signals generated by the first and second working electrodes were processed to provide the data shown in **Figs. 17A-17B.** The glucose values provided by the dual-electrode sensor tracked very closely with the control measurements provided by the YSI device. Additionally, all glucose values provided by the dual-electrode glucose sensor fell within the "A" portion of the Clark Error Grid. The sensor had an MARD of 7.5% and met the FDA's ISO standard for hand-held glucose monitors (±15-mg/dl or ±20%) by 100%.

**[0916]** From these data, the investigators concluded that continuous IV glucose tracking (*via* a peripheral vein) using a dual-electrode continuous glucose sensor, including first and second working electrodes configured as described herein, coupled with a catheter and a flow control system can provide consistently accurate glucose values *in vivo.*

**[0917]** Methods and devices that are suitable for use in conjunction with aspects of the preferred embodiments are disclosed in U.S. Patent No. 4,994,167; U.S. Patent No. 4,757,022; U.S. Patent No. 6,001,067; U.S. Patent No. 6,741,877; U.S. Patent No. 6,702,857; U.S. Patent No. 6,558,321; U.S. Patent No. 6,931,327; U.S. Patent No. 6,862,465; U.S. Patent No. 7,074,307; U.S. Patent No. 7,081,195; U.S. Patent No. 7,108,778; U.S. Patent No. 7,110,803; U.S. Patent No. 7,192,450; U.S. Patent No. 7,226,978; and U.S. Patent No. 7,310,544.

**[0918]** Methods and devices that are suitable for use in conjunction with aspects of the preferred embodiments are disclosed in U.S. Patent Publication No. US-2005-0176136-A1; U.S. Patent Publication No. US-2005-0251083-A1; U.S. Patent Publication No. US-2005-0143635-A1; U.S. Patent Publication No. US-2005-0181012-A1; U.S. Patent Publication No. US-2005-0177036-A1; U.S. Patent Publication No. US-2005-0124873-A1; U.S. Patent Publication No. US-2005-0115832-A1; U.S. Patent Publication No. US-2005-0245799-A1; U.S. Patent Publication No. US-2005-0245795-A1; U.S. Patent Publication No. US-2005-0242479-A1; U.S. Patent Publication No. US-2005-0182451-A1; U.S. Patent Publication No. US-2005-0056552-A1; U.S. Patent Publication No. US-2005-0192557-A1; U.S. Patent Publication No. US-2005-0154271-A1; U.S. Patent Publication No. US-2004-0199059-A1; U.S. Patent Publication No. US-2005-0054909-A1; U.S. Patent Publication No. US-2005-0051427-Al; U.S. Patent Publication No. US-2003-0032874-Al; U.S. Patent Publication No. US-2005-0103625-Al; U.S. Patent Publication No. US-2005-0203360-A1; U.S. Patent Publication No. US-2005-0090607-A1; U.S. Patent Publication No. US-2005-0187720-A1; U.S. Patent Publication No. US-2005-0161346-A1; U.S. Patent Publication No. US-2006-0015020-A1; U.S. Patent Publication No. US-2005-0043598-A1; U.S. Patent Publication No. US-2005-0033132-A1; U.S. Patent Publication No. US-2005-0031689-A1; U.S. Patent Publication No. US-2004-0186362-A1; U.S. Patent Publication No. US-2005-0027463-A1; U.S. Patent Publication No. US-2005-0027181-A1; U.S. Patent Publication No. US-2005-0027180-A1; U.S. Patent Publication No. US-2006-0020187-A1; U.S. Patent Publication No. US-2006-0036142-A1; U.S. Patent Publication No. US-2006-0020192-A1; U.S. Patent Publication No. US-2006-0036143-A1; U.S. Patent Publication No. US-2006-0036140-A1; U.S. Patent Publication No. US-2006-0019327-A1; U.S. Patent Publication No. US-2006-0020186-A1; U.S. Patent Publication No. US-2006-0036139-A1; U.S. Patent Publication No. US-2006-0020191-A1; U.S. Patent Publication No. US-2006-0020188-A1; U.S. Patent Publication No. US-2006-0036141-A1; U.S. Patent Publication No. US-

2006-0020190-A1; U.S. Patent Publication No. US-2006-0036145-A1; U.S. Patent Publication No. US-2006-0036144-A1; U.S. Patent Publication No. US-2006-0016700-A1; U.S. Patent Publication No. US-2006-0142651-A1; U.S. Patent Publication No. US-2006-0086624-A1; U.S. Patent Publication No. US-2006-0068208-A1; U.S. Patent Publication No. US-2006-0040402-A1; U.S. Patent Publication No. US-2006-0036142-A1; U.S. Patent Publication No. US-2006-0036141-A1; U.S. Patent Publication No. US-2006-0036143-A1; U.S. Patent Publication No. US-2006-0036140-A1; U.S. Patent Publication No. US-2006-0036139-A1; U.S. Patent Publication No. US-2006-0142651-A1; U.S. Patent Publication No. US-2006-0036145-A1; U.S. Patent Publication No. US-2006-0036144-A1; U.S. Patent Publication No. US-2006-0200022-A1; U.S. Patent Publication No. US-2006-0198864-A1; U.S. Patent Publication No. US-2006-0200019-A1; U.S. Patent Publication No. US-2006-0189856-A1; U.S. Patent Publication No. US-2006-0200020-A1; U.S. Patent Publication No. US-2006-0200970-A1; U.S. Patent Publication No. US-2006-0183984-A1; U.S. Patent Publication No. US-2006-0183985-A1; U.S. Patent Publication No. US-2006-0195029-A1; U.S. Patent Publication No. US-2006-0229512-A1; U.S. Patent Publication No. US-2006-0222566-A1; U.S. Patent Publication No. US-2007-0032706-A1; U.S. Patent Publication No. US-2007-0016381-A1; U.S. Patent Publication No. US-2007-0027370-A1; U.S. Patent Publication No. US-2007-0027384-A1; U.S. Patent Publication No. US-2007-0032717-A1; U.S. Patent Publication No. US-2007-0032718-A1; U.S. Patent Publication No. US-2007-0059196-A1; U.S. Patent Publication No. US-2007-0066873-A1; U.S. Patent Publication No. US-2007-0093704-A1; U.S. Patent Publication No. US-2007-0197890-A1; U.S. Patent Publication No. US-2007-0173710-A1; U.S. Patent Publication No. US-2007-0163880-A1; U.S. Patent Publication No. US-2007-0203966-A1; U.S. Patent Publication No. US-2007-0213611-A1; U.S. Patent Publication No. US-2007-0232879-A1; U.S. Patent Publication No. US-2007-0235331-A1; U.S. Patent Publication No. US-2008-0021666-A1; and U.S. Patent Publication No. US-2008-0033254-A1.

**[0919]** Methods and devices that are suitable for use in conjunction with aspects of the preferred embodiments are disclosed in U.S. Patent Application No. 09/447,227 filed November 22, 1999 and entitled "DEVICE AND METHOD FOR DETERMINING ANALYTE LEVELS"; U.S. Patent Application No. 11/654,135 filed January 17, 2007 and entitled "POROUS MEMBRANES FOR USE WITH IMPLANTABLE DEVICES"; U.S. Patent Application No. 11/654,140 filed January 17, 2007 and entitled "MEMBRANES FOR AN ANALYTE SENSOR"; U.S. Patent Application No. 11/543,396 filed October 4, 2006 and entitled "ANALYTE SENSOR"; U.S. Patent Application No. 11/543,490 filed October 4, 2006 and entitled "ANALYTE SENSOR"; U.S. Patent Application No. 11/543,404 filed October 4, 2006 and entitled "ANALYTE SENSOR"; U.S. Patent Application No. 11/691,426 filed March 26, 2007 and entitled "ANALYTE SENSOR"; U.S. Patent Application No. 11/691,432 filed March 26, 2007 and entitled "ANALYTE SENSOR"; U.S. Patent Application No. 11/691,424 filed March 26, 2007 and entitled "ANALYTE SENSOR"; U.S. Patent Application No. 11/691,466 filed March 26, 2007 and entitled "ANALYTE SENSOR"; U.S. Patent Application No. 12/037,830 filed February 26, 2008 and entitled "ANALYTE MEASURING DEVICE"; and U.S. Patent Application No. 12/037,812 filed February 26,2008 and entitled "ANALYTE MEASURING DEVICE".

**[0920]** The term "comprising" as used herein is synonymous with "including," "containing," or "characterized by," and is inclusive or open-ended and does not exclude additional, unrecited elements or method steps.

**[0921]** All numbers expressing quantities of ingredients, reaction conditions, and so forth used in the specification are to be understood as being modified in all instances by the term "about." Accordingly, unless indicated to the contrary, the numerical parameters set forth herein are approximations that may vary depending upon the desired properties sought to be obtained. At the very least, and not as an attempt to limit the application of the doctrine of equivalents to the scope of any claims in any application claiming priority to the present application, each numerical parameter should be construed in light of the number of significant digits and ordinary rounding approaches.

**[0922]** The above description discloses several methods and materials of the present invention. This invention is susceptible to modifications in the methods and materials, as well as alterations in the fabrication methods and equipment. Such modifications will become apparent to those skilled in the art from a consideration of this disclosure or practice of the invention disclosed herein. Consequently, it is not intended that this invention be limited to the specific embodiments disclosed herein, but that it cover all modifications and alternatives coming within the scope of the invention.

## Claims

**1.** A system (600) for continuous measurement of a glucose concentration, the system comprising:

a continuous glucose sensor (14) comprising a first working electrode (E1) and a second working electrode (E2), the first working electrode disposed beneath an active enzymatic portion of a sensor membrane (347, 348, 349, 350) and configured to generate a first signal associated with both glucose and non-glucose related electroactive compounds, the second working electrode disposed beneath an inactive-enzymatic or a non-enzymatic portion of the sensor membrane and configured to generate a non-glucose-related signal associated with the non-glucose electroactive species; wherein the continuous glucose sensor is located at at least one of

in a vascular access device (12) and on a vascular access device configured for fluid communication with a host's circulatory system and the first working electrode and the second working electrode configured to provide one or more sensor data points; and

a flow control device (604) configured to intermittently meter a glucose reference solution (602a) across the continuous glucose sensor such that the continuous glucose sensor intermittently measures the glucose concentration of the reference solution and to provide one or more reference glucose data points,

a processor module configured to match the one or more sensor data points with the one or more reference glucose data point to form one or more substantially time-corresponding matched data pairs and create a calibration set; and

wherein the system is configured to calibrate the continuous glucose sensor using the measured glucose concentration of the reference solution and the calibration set.

**2.** The system (600) of Claim 1, wherein the processor module is configured to process signals from the first working electrode (E1) and second working electrode (E2) to obtain a substantially baseline-free glucose signal by subtracting the second signal from the first signal.

**3.** The system (600) of Claim 2, wherein the processor module is configured to calibrate the continuous glucose sensor (14) using a regression equation in combination with the calibration set that does not include a baseline parameter.

**4.** The system (600) of Claim 1, wherein the processor module is configured to use the measured glucose concentration of the reference solution (602a) to determine a sensitivity of the continuous glucose sensor (14).

**5.** The system (600) of Claim 4, wherein the processor module is configured to calibrate the continuous glucose sensor (14) using the sensitivity of the continuous glucose sensor.

**6.** The system (600) of Claim 1, wherein the processor module is configured to auto-calibrate the sensor without an external reference value.

**7.** The system (600) of claim 1, wherein the flow control device (604) is configured to maintain the continuous glucose sensor (14) in contact with the glucose reference solution (602a) at least about 50% of the time during a sensor session.

**8.** The system (600) of Claim 7, wherein the flow control device (604) is configured to meter the reference solution (602a) such that the sensor is contacting the reference solution at least about 65% of the time during a sensor session.

**9.** The system (600) of Claim 7, wherein the flow control device (604) is configured to meter the reference solution (602a) such that the sensor is contacting the reference solution at least about 80% of the time during a sensor session.

**10.** The system (600) of Claim 1, wherein the flow control device (604) is configured to meter the reference solution (602a) such that the sensor is contacting the reference solution a sufficient amount of time such that biofouling does not occur for at least about 3 days of sensor use.

**11.** The system (600) of Claim 10, wherein the sufficient amount of time is at least about 50%, or at least about 65%, of the time during sensor use.

**12.** The system (600) of Claim 10, wherein the sufficient amount of time is at least about 80% of the time during sensor use.

**13.** The system (600) of Claim 10, wherein the flow control device (604) is configured to meter the reference solution (602a) through the vascular access device (12) for a time and with a sufficient flow rate such that the vascular access device remains patent during a sensor session of at least about 3 days.

**14.** The system (600) of Claim 13, wherein the flow control device (604) is configured to meter the reference solution (602a) through the vascular access device (12) for at least about 50% of a sensor session, at a flow rate from about 0.001 ml/min to about 2.0 ml/min.

**15.** The system (600) of Claim 13, wherein the flow control device (604) is configured to meter the reference solution (602a) through the vascular access device (12) for at least about 65% of a sensor session, at a flow rate from about 0.5 ml/min to about 2.0 ml/min, whereby the vascular access device remains patent during a sensor session of from

about 3 days to about 30 days.

**Patentansprüche**

1. System (600) zur kontinuierlichen Messung einer Glukosekonzentration, wobei das System umfasst:

einen kontinuierlichen Glukosesensor (14) umfassend eine erste Arbeitselektrode (E1) und eine zweite Arbeitselektrode (E2), die erste Arbeitselektrode angeordnet unter einem aktiven enzymatischen Abschnitt einer Sensormembran (347, 348, 349, 350) und
konfiguriert zur Erzeugung eines ersten Signals, das sowohl mit Glukose als auch mit nicht-Glukose-bezogenen elektroaktiven Verbindungen assoziiert ist, die zweite Arbeitselektrode angeordnet unter einem inaktiven enzymatischen oder einem nichtenzymatischen Abschnitt der Sensormembran und konfiguriert zur Erzeugung eines nicht-Glukose-bezogenen Signals, das mit einer elektroaktiven nicht-Glukose-Spezies assoziiert ist; wobei der kontinuierliche Glukosesensor zumindest in einer Gefäßzugangsvorrichtung (12) und an einer Gefäßzugangsvorrichtung angeordnet ist, die zur Fluidkommunikation mit einem Kreislaufsystem eines Wirts konfiguriert ist und die erste Arbeitselektrode und die zweite Arbeitselektrode konfiguriert sind, um einen oder mehrere Sensordatenpunkte bereitzustellen; und
eine Vorrichtung zur Flusssteuerung (604), die konfiguriert ist, um intermittierend eine Glukosereferenzlösung (602a) über den kontinuierlichen Glukosesensor abzugeben,
derart dass der kontinuierliche Glukosesensor intermittierend die Glukosekonzentration der Referenzlösung misst und um einen oder mehrere Referenz-Glukose-Datenpunkte bereitzustellen,
ein Prozessormodul, das konfiguriert ist, um den einen oder mehrere Sensordatenpunkte mit dem einen oder mehreren Referenz-Glukose-Datenpunkt abzugleichen, um ein oder mehrere im Wesentlichen zeitlich korrespondierende, zusammenpassende Datenpaare zu bilden, und einen Kalibrierungssatz zu erstellen; und
wobei das System konfiguriert ist, den kontinuierlichen Glukosesensor unter Verwendung der gemessenen Glukosekonzentration der Referenzlösung und des Kalibrierungssatzes zu kalibrieren.

2. System (600) nach Anspruch 1, wobei das Prozessormodul konfiguriert ist, Signale von der ersten Arbeitselektrode (E1) und der zweiten Arbeitselektrode (E2) zu verarbeiten, um ein im wesentlichen grundlinienfreies Glukosesignal zu erhalten durch Subtrahieren des zweiten Signals von dem ersten Signal.

3. System (600) nach Anspruch 2, wobei das Prozessormodul konfiguriert ist, den kontinuierlichen Glukosesensor (14) unter Verwendung einer Regressionsgleichung in Verbindung mit dem Kalibrierungssatz, der einen Grundlinien-Parameter nicht enthält, zu kalibrieren.

4. System (600) nach Anspruch 1, wobei das Prozessormodul konfiguriert ist, die gemessene Glukose-Konzentration der Referenzlösung (602a) zu verwenden, um eine Empfindlichkeit des kontinuierlichen Glukosesensors (14) zu bestimmen.

5. System (600) nach Anspruch 4, wobei das Prozessormodul konfiguriert ist, den kontinuierlichen Glukosesensor (14) unter Verwendung der Empfindlichkeit des kontinuierlichen Glukosesensors zu kalibrieren.

6. System (600) nach Anspruch 1, wobei das Prozessormodul konfiguriert ist, den Sensor ohne einen externen Referenzwert zu autokalibrieren.

7. System (600) nach Anspruch 1, wobei die Vorrichtung zur Flusssteuerung (604) konfiguriert ist, den kontinuierlichen Glukosesensor (14) in Kontakt mit der Glukosereferenzlösung (602a) für mindestens etwa 50% der Zeit während der Sensorsitzung zu halten.

8. System (600) nach Anspruch 7, wobei die Vorrichtung zur Flusssteuerung (604) konfiguriert ist die Referenzlösung (602a) derart abzugeben, dass der Sensor während einer Sensorsitzung für mindestens etwa 65% der Zeit mit der Referenzlösung in Kontakt ist.

9. System (600) nach Anspruch 7, wobei die Vorrichtung zur Flusssteuerung (604) konfiguriert ist die Referenzlösung (602a) derart abzugeben, dass der Sensor während einer Sensorsitzung für mindestens etwa 80% der Zeit mit der Referenzlösung in Kontakt ist.

**10.** System (600) nach Anspruch 1, wobei die Vorrichtung zur Flusssteuerung (604) konfiguriert ist die Referenzlösung (602a) derart abzugeben, dass der Sensor für eine ausreichende Zeitdauer in Kontakt mit der Referenzlösung ist, so dass Biofouling für mindestens etwa 3 Tage der Sensorverwendung nicht auftritt.

**11.** System (600) nach Anspruch 10, wobei die ausreichende Zeitdauer mindestens etwa 50% oder mindestens etwa 65% der Zeit während der Sensorverwendung beträgt.

**12.** System (600) nach Anspruch 10, wobei die ausreichende Zeitdauer mindestens etwa 80% der Zeit während der Sensorverwendung beträgt.

**13.** System (600) nach Anspruch 10, wobei die Vorrichtung zur Flusssteuerung (604) konfiguriert ist zur Abgabe der Referenzlösung (602a) durch die Gefäßzugangsvorrichtung (12) für eine Zeitdauer und mit einer ausreichenden Flussrate, derart dass die Gefäßzugangsvorrichtung offen verbleibt während einer Sensorsitzung von mindestens ca. 3 Tagen.

**14.** System (600) nach Anspruch 13, wobei die Vorrichtung zur Flusssteuerung (604) konfiguriert ist zur Abgabe der Referenzlösung (602a) durch die Gefäßzugangsvorrichtung (12) für mindestens etwa 50% einer Sensorsitzung bei einer Flussrate von etwa 0,001 ml/min bis etwa 2,0 ml/min.

**15.** System (600) nach Anspruch 13, wobei die Vorrichtung zur Flusssteuerung (604) konfiguriert ist zur Abgabe der Referenzlösung (602a) durch die Gefäßzugangsvorrichtung (12) für mindestens etwa 65% einer Sensorsitzung bei einer Flussrate von etwa 0,5 ml/min bis etwa 2,0 ml/min, wobei die Gefäßzugangsvorrichtung offen verbleibt während eines Sensorsitzung von etwa 3 Tagen bis etwa 30 Tagen.

**Revendications**

**1.** Système (600) pour la mesure continue d'une concentration de glucose, le système comprenant :

un capteur de glucose continu (14) comprenant une première électrode de travail (E1) et une deuxième électrode de travail (E2), la première électrode de travail étant disposée au-dessous d'une partie enzymatique active d'une membrane du capteur (347, 348, 349, 350) et configurée pour générer un premier signal associé aux composés électroactifs relatifs à la fois au glucose et au non glucose, la deuxième électrode de travail étant disposée au-dessous d'une partie enzymatique inactive ou d'une partie non enzymatique de la membrane du capteur et configurée pour générer un signal relatif au non glucose associé aux espèces électroactives du non glucose ; dans lequel le capteur de glucose continu est situé au moins dans un dispositif d'accès vasculaire (12) et/ou sur un dispositif d'accès vasculaire configuré pour une communication fluidique avec un système circulatoire d'hôte et la première électrode de travail et la deuxième électrode de travail sont configurées pour fournir un ou plusieurs points de données de capteur ; et

un dispositif de commande d'écoulement (604) configuré pour mesurer par intermittence une solution de référence de glucose (602a) à travers le capteur de glucose continu de sorte que le capteur de glucose continu mesure par intermittence la concentration de glucose de la solution de référence et pour fournir un ou plusieurs points de données de glucose de référence,

un module de processeur configuré pour mettre en correspondance lesdits un ou plusieurs points de données de capteur avec lesdits un ou plusieurs points de données de glucose de référence pour former une ou plusieurs paires de données correspondantes sensiblement correspondantes dans le temps et créer un ensemble d'étalonnage ; et

dans lequel le système est configuré pour étalonner le capteur de glucose continu en utilisant la concentration de glucose mesurée de la solution de référence et l'ensemble d'étalonnage.

**2.** Système (600) selon la revendication 1, dans lequel le module de processeur est configuré pour traiter les signaux provenant de la première électrode de travail (E1) et de la deuxième électrode de travail (E2) pour obtenir un signal de glucose sensiblement exempt de ligne de base en soustrayant le deuxième signal du premier signal.

**3.** Système (600) selon la revendication 2, dans lequel le module de processeur est configuré pour étalonner le capteur de glucose continu (14) en utilisant une équation de régression en combinaison avec l'ensemble d'étalonnage qui ne comprend pas de paramètre de ligne de base.

**4.** Système (600) selon la revendication 1, dans lequel le module de processeur est configuré pour utiliser la concentration de glucose mesurée de la solution de référence (602a) pour déterminer une sensibilité du capteur de glucose continu (14).

**5.** Système (600) selon la revendication 4, dans lequel le module de processeur est configuré pour étalonner le capteur de glucose continu (14) en utilisant la sensibilité du capteur de glucose continu.

**6.** Système (600) selon la revendication 1, dans lequel le module de processeur est configuré pour auto-étalonner le capteur sans valeur de référence externe.

**7.** Système (600) selon la revendication 1, dans lequel le dispositif de commande d'écoulement (604) est configuré pour maintenir le capteur de glucose continu (14) en contact avec la solution de référence de glucose (602a) pendant au moins environ 50 % du temps d'une session de capteur.

**8.** Système (600) selon la revendication 7, dans lequel le dispositif de commande d'écoulement (604) est configuré pour mesurer la solution de référence (602a) de sorte que le capteur soit en contact avec la solution de référence pendant au moins environ 65 % du temps d'une session de capteur.

**9.** Système (600) selon la revendication 7, dans lequel le dispositif de commande d'écoulement (604) est configuré pour mesurer la solution de référence (602a) de sorte que le capteur soit en contact avec la solution de référence pendant au moins environ 80 % du temps d'une session de capteur.

**10.** Système (600) selon la revendication 1, dans lequel le dispositif de commande d'écoulement (604) est configuré pour mesurer la solution de référence (602a) de sorte que le capteur soit en contact avec la solution de référence pendant une quantité de temps suffisante pour qu'un encrassement biologique ne se produise pas pendant au moins environ 3 jours d'utilisation du capteur.

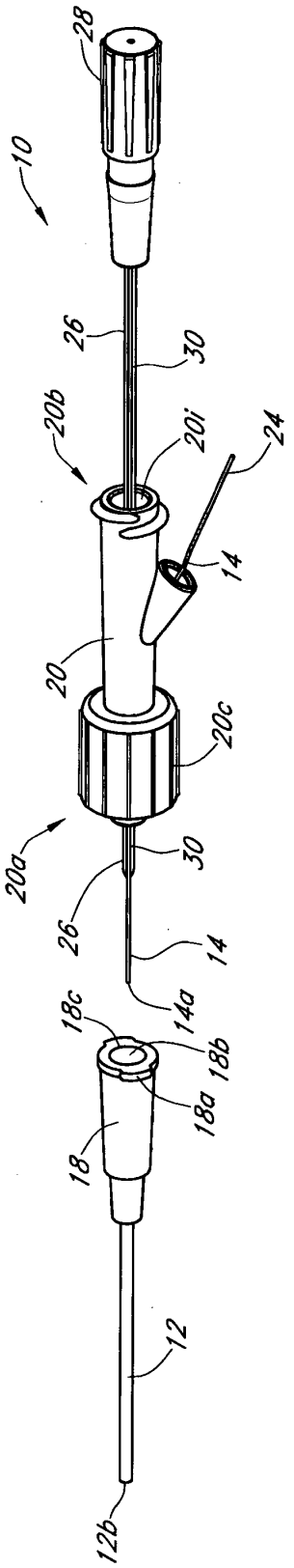
**11.** Système (600) selon la revendication 10, dans lequel la quantité de temps suffisante est d'au moins environ 50 %, ou d'au moins environ 65 %, du temps d'utilisation du capteur.

**12.** Système (600) selon la revendication 10, dans lequel la quantité de temps suffisante est d'au moins environ 80 % du temps d'utilisation du capteur.

**13.** Système (600) selon la revendication 10, dans lequel le dispositif de commande d'écoulement (604) est configuré pour mesurer la solution de référence (602a) par le dispositif d'accès vasculaire (12) pendant un temps et avec un débit suffisant pour que le dispositif d'accès vasculaire reste patent pendant une session de capteur d'au moins environ 3 jours.

**14.** Système (600) selon la revendication 13, dans lequel le dispositif de commande d'écoulement (604) est configuré pour mesurer la solution de référence (602a) par le dispositif d'accès vasculaire (12) pendant au moins environ 50 % d'une session de capteur, à un débit d'environ 0,001 ml/mn à environ 2,0 ml/mn.

**15.** Système (600) selon la revendication 13, dans lequel le dispositif de commande d'écoulement (604) est configuré pour mesurer la solution de référence (602a) par le dispositif d'accès vasculaire (12) pendant au moins environ 65 % d'une session de capteur, à un débit d'environ 0,5 ml/mn à environ 2,0 ml/m, moyennant quoi le dispositif d'accès vasculaire reste patent pendant une session de capteur d'environ 3 jours à environ 30 jours.


10
28
26
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20i
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20c
20a
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14
14a
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18b
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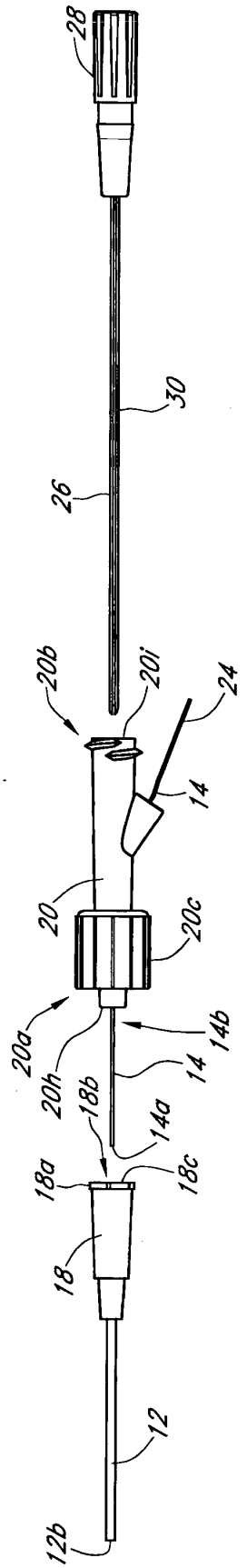
18
12
12b

FIG. 1A


28
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26
20b
20i
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14
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20c
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18c
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12
12b

FIG. 1B

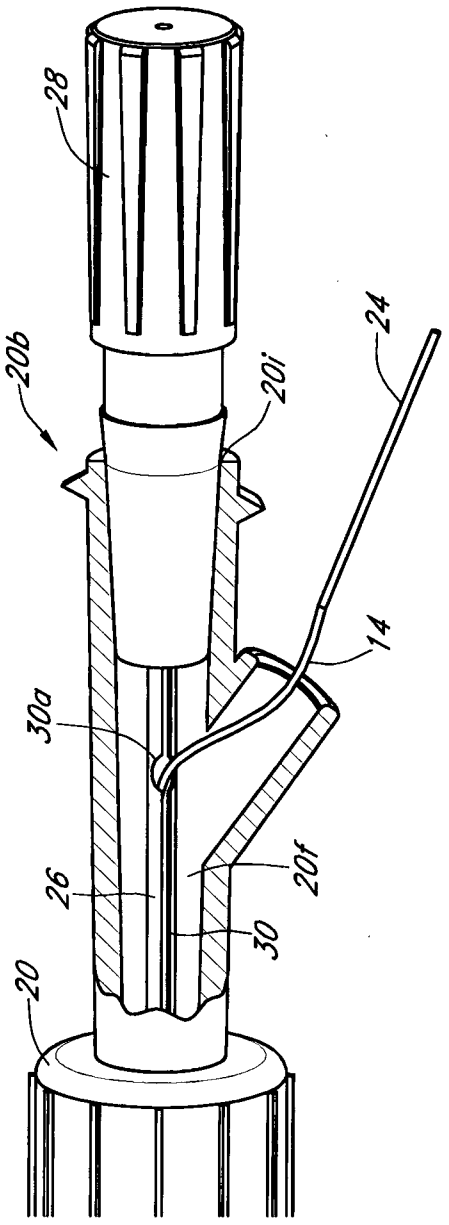


FIG. 1C$_1$

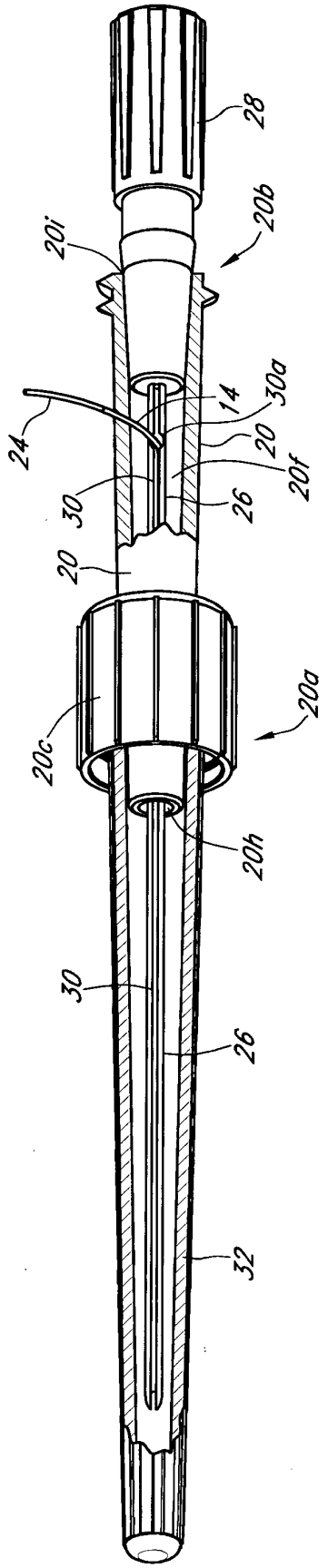


FIG. 1C$_2$

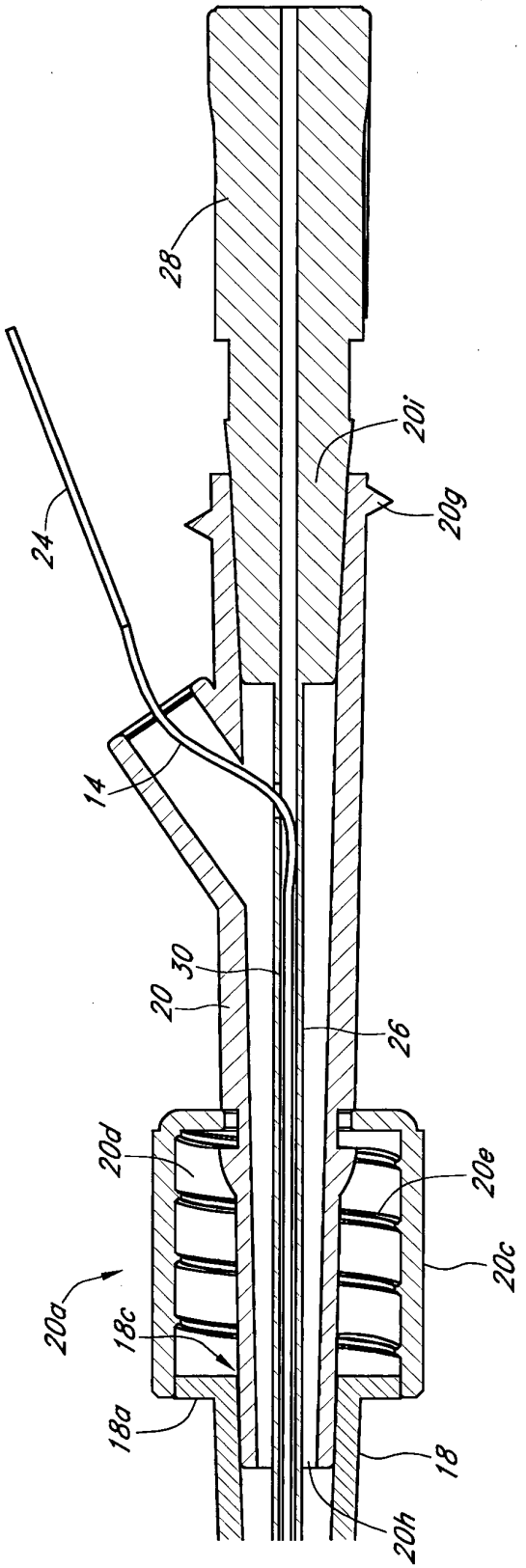
28
20i
20g
24
14
30
20
26
20d
20e
20c
20a
18c
18a
18
20h

FIG. 1D

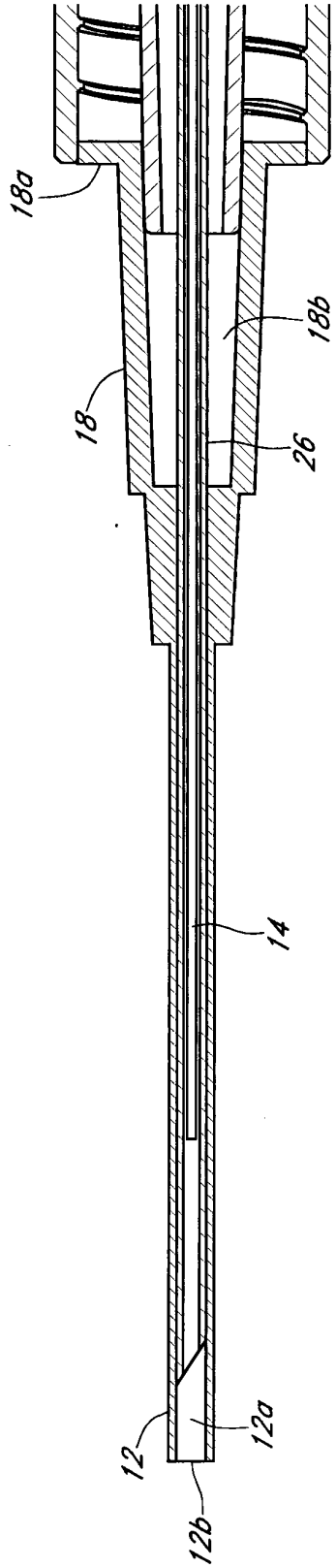
18a
18b
18
26
14
12
12a
12b

FIG. 1E

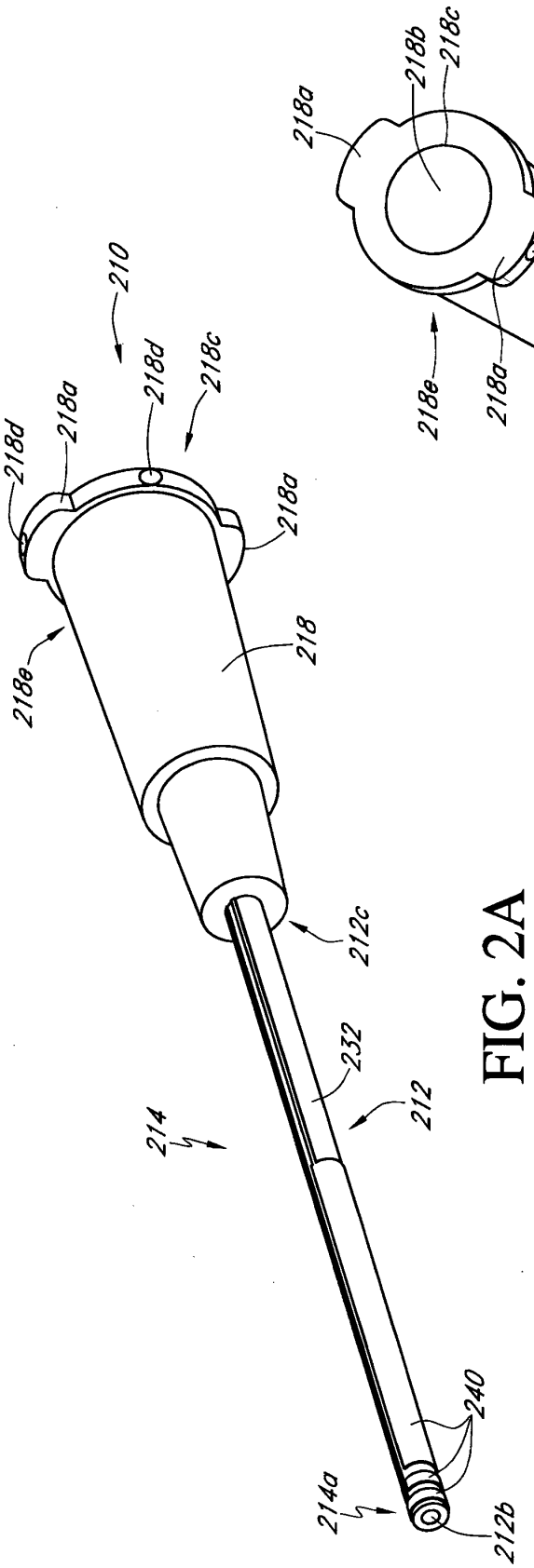
210
218d
218a
218c
218d
218a
218a
218
218e
212c
232
212
214
240
214a
212b

FIG. 2A

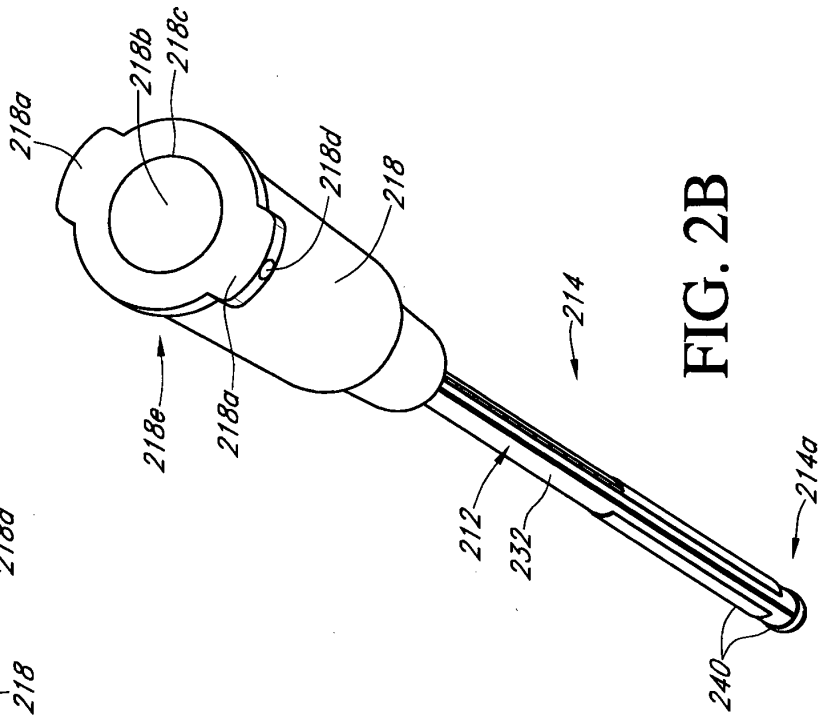
218a
218b
218c
218d
218
214
218e
218a
212
232
214a
240

FIG. 2B

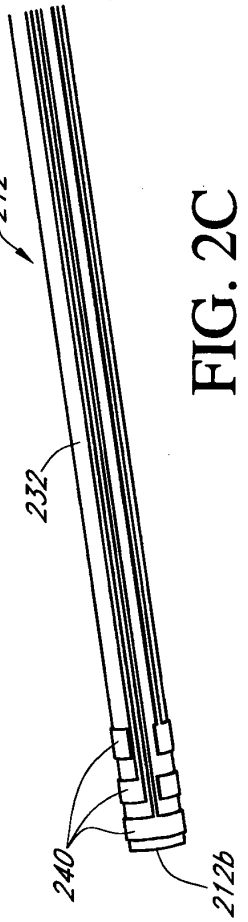


FIG. 2C

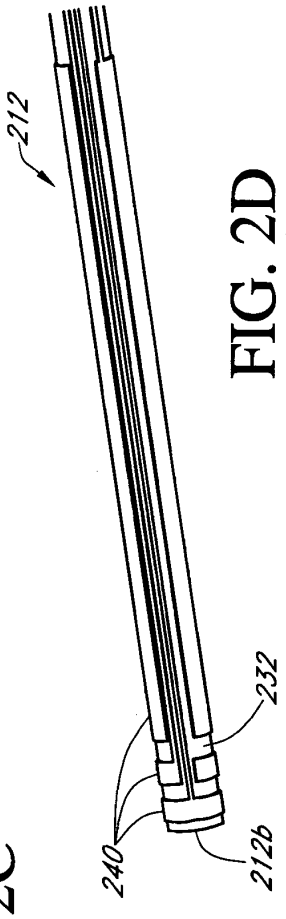


FIG. 2D

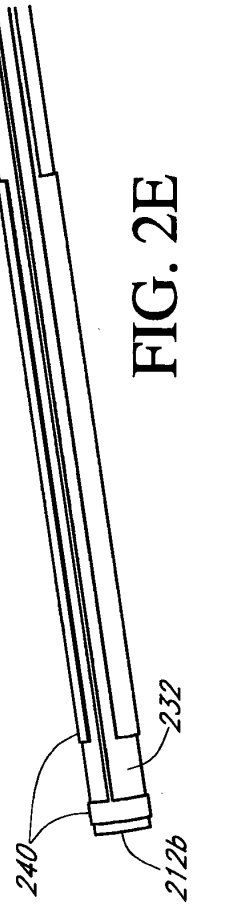


FIG. 2E

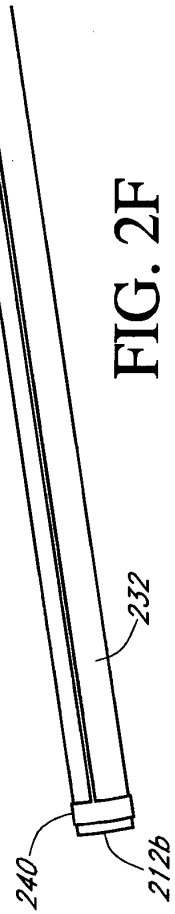


FIG. 2F

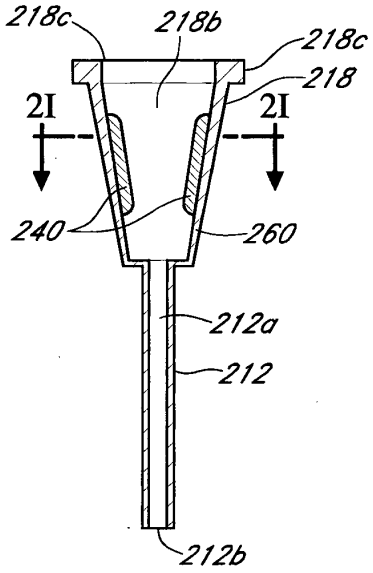


FIG. 2G

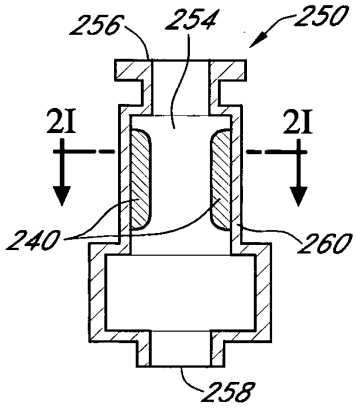


FIG. 2H

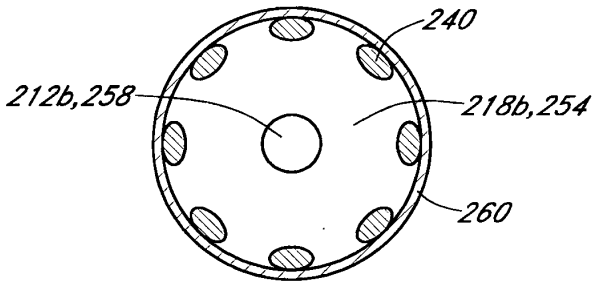


FIG. 2I

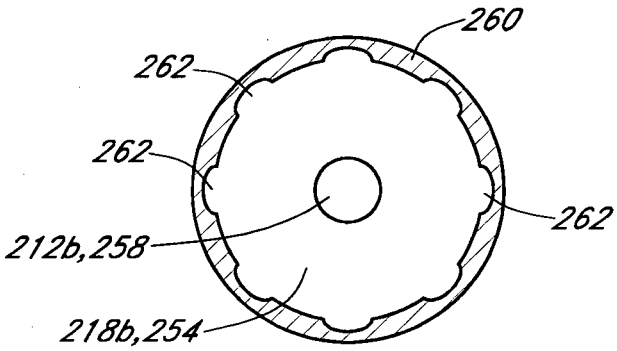


FIG. 2J

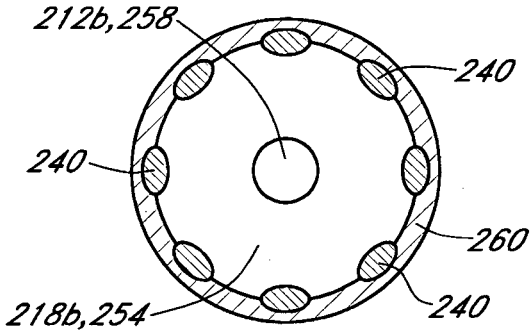


FIG. 2K

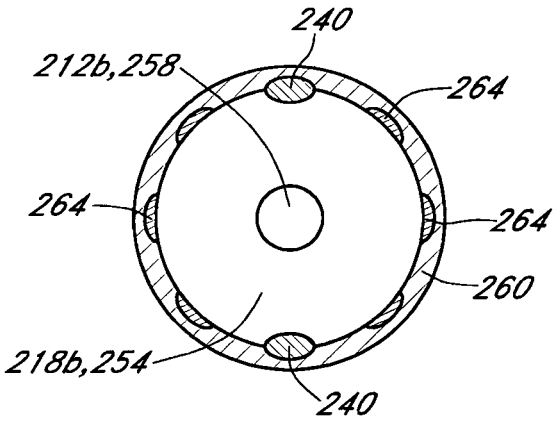


FIG. 2L

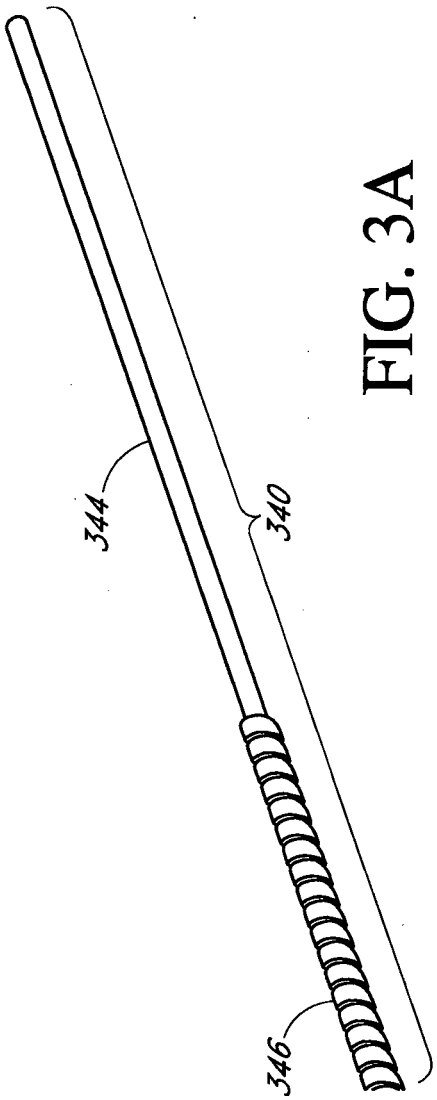


FIG. 3A

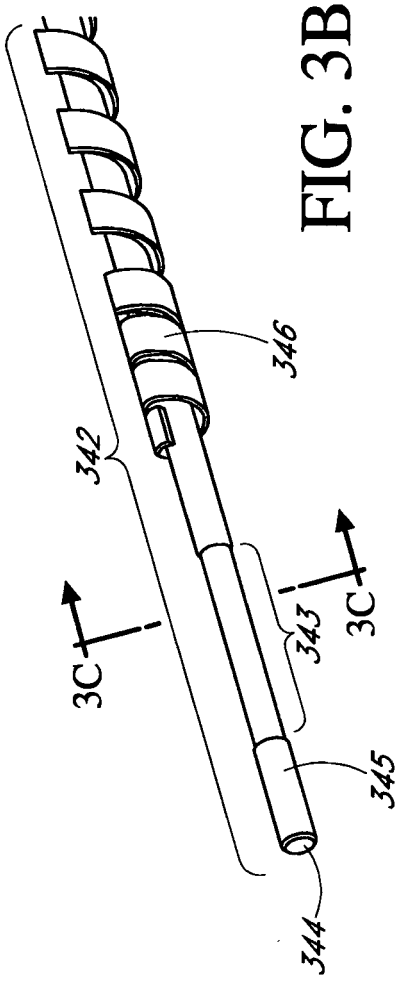


FIG. 3B

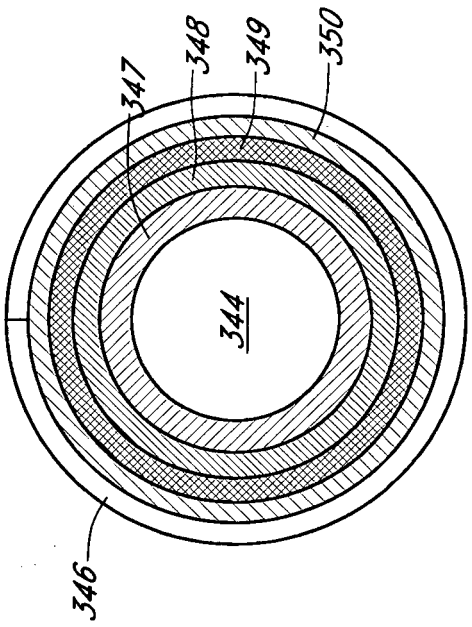


FIG. 3C

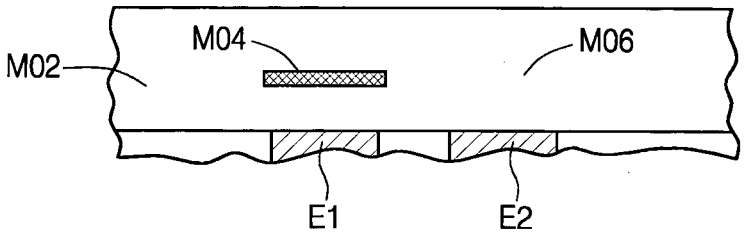


FIG. 3D

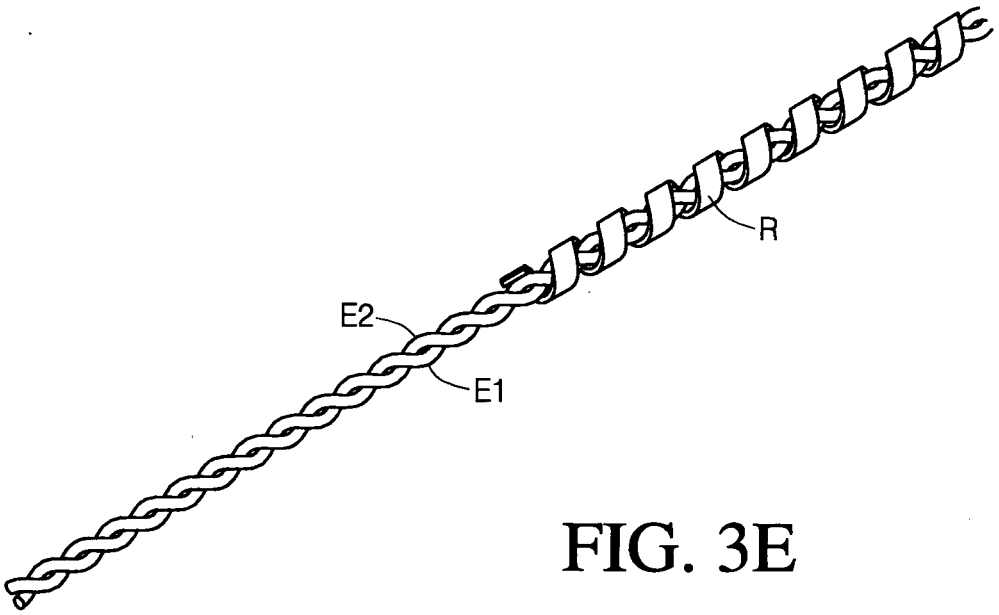


FIG. 3E

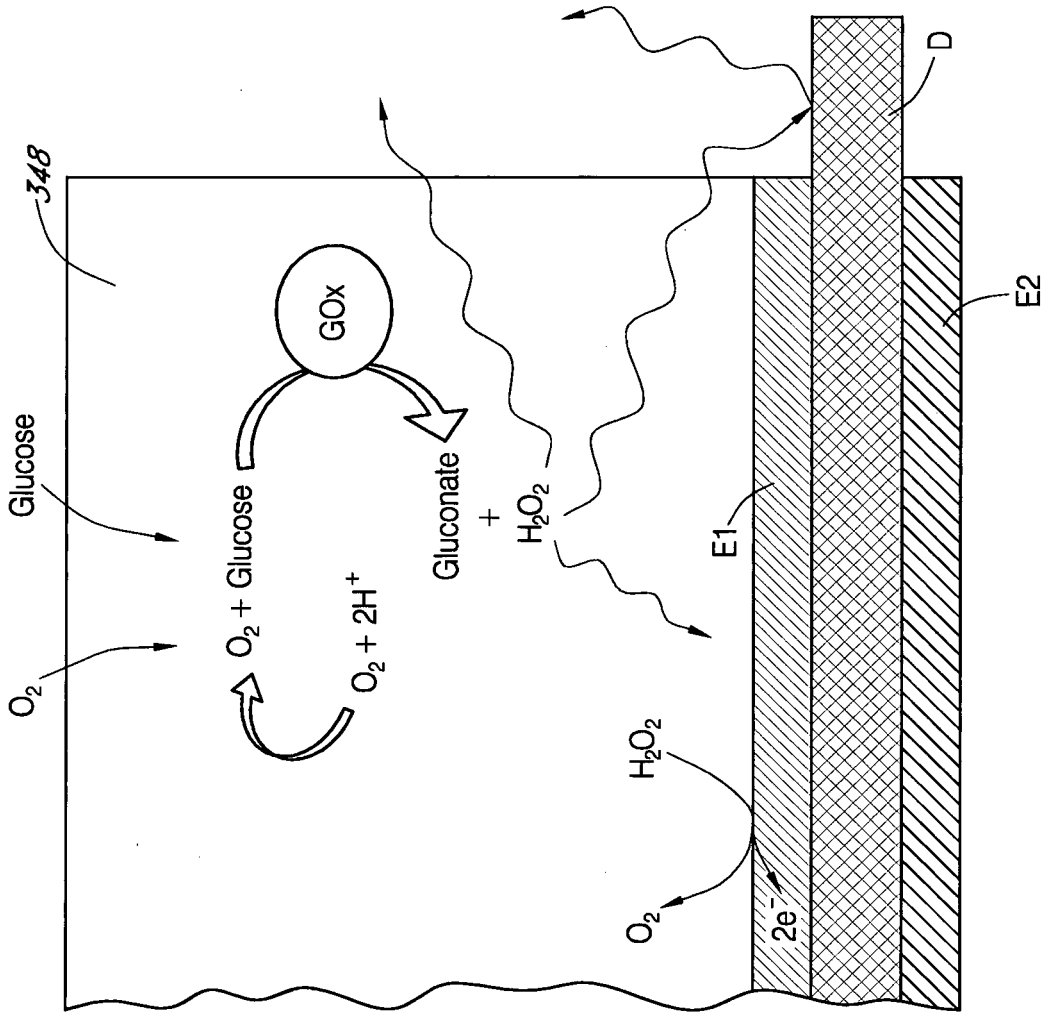
348
Glucose
$O_2$
GOx
$O_2$ + Glucose
$O_2$ + $2H^+$
Gluconate
+
$H_2O_2$
D
E2
E1
$H_2O_2$
$O_2$
$2e^-$

FIG. 3F

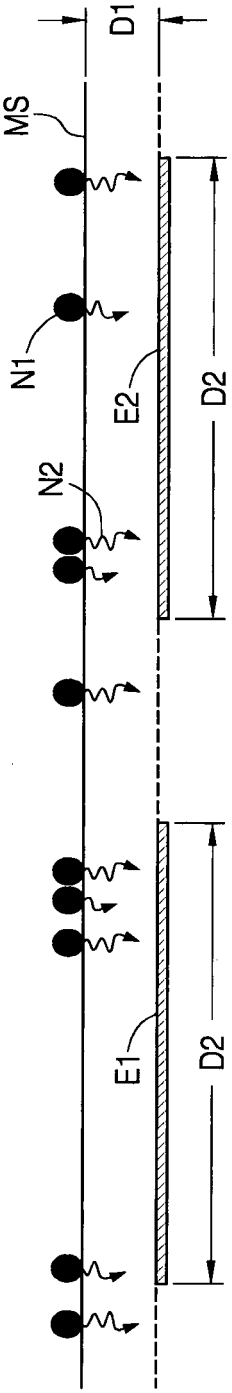


FIG. 3G

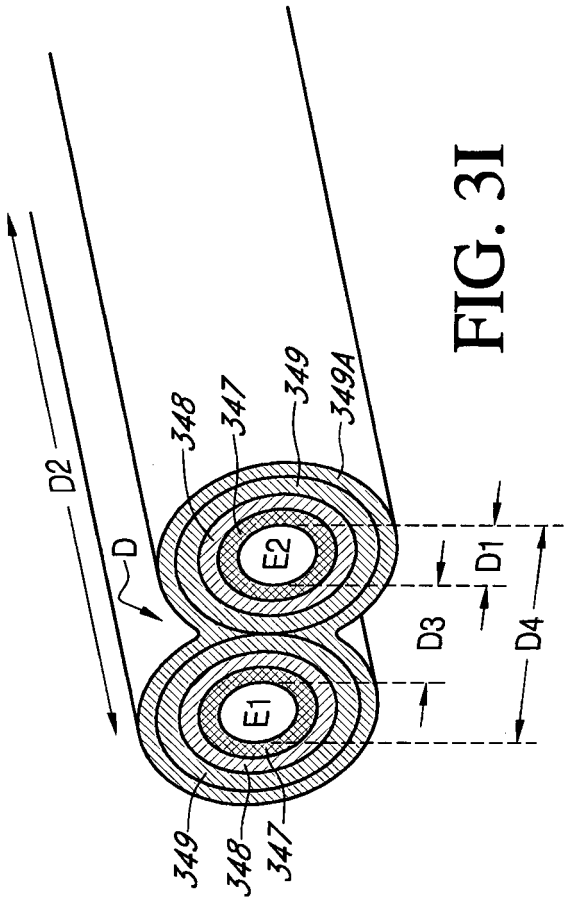


FIG. 3I

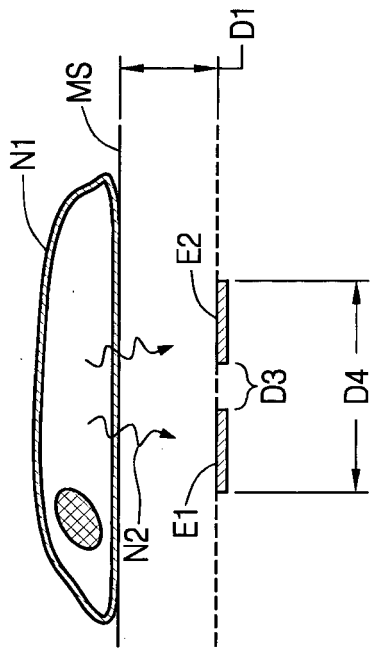


FIG. 3H

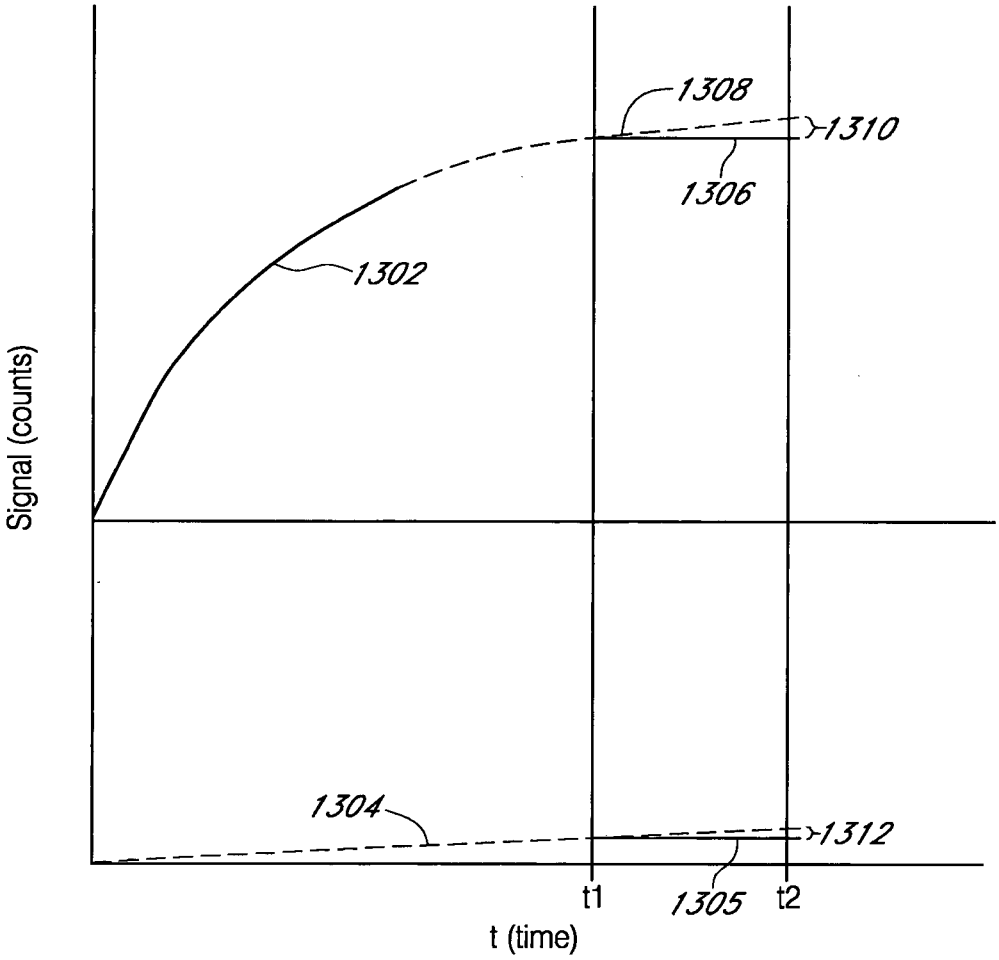


FIG. 3J

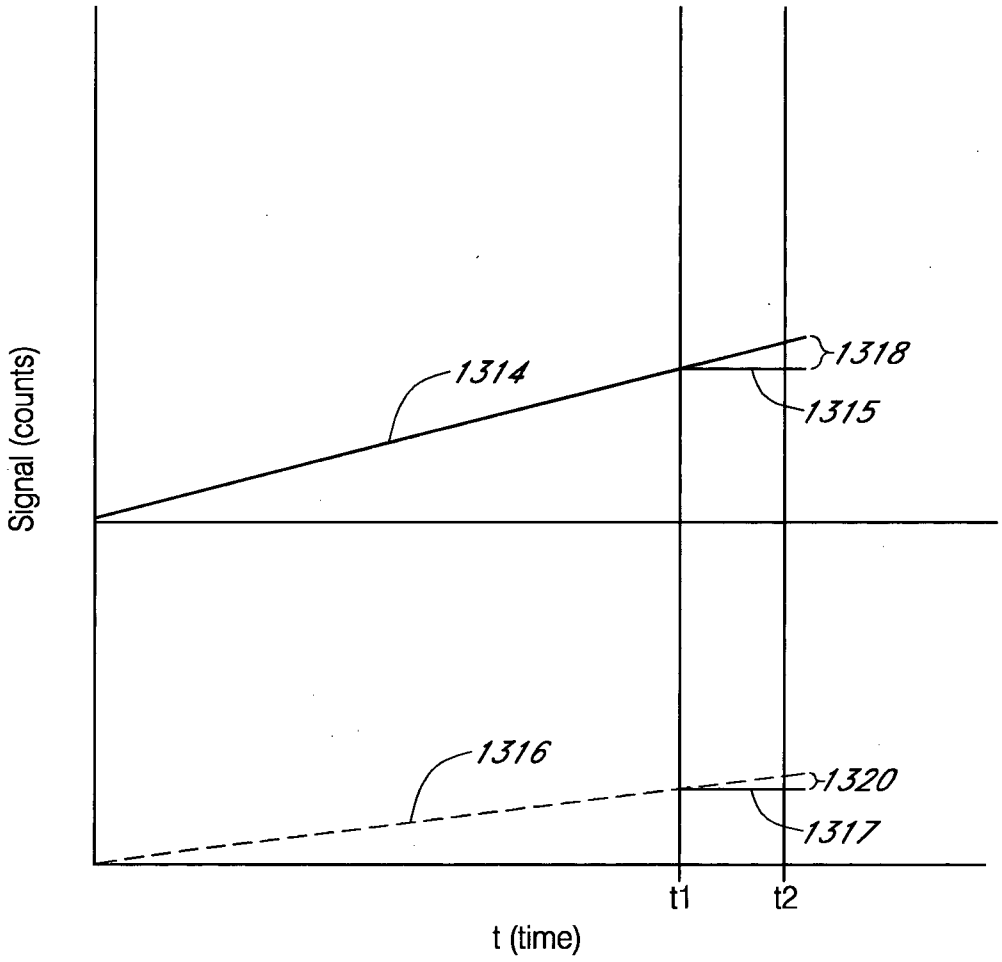


FIG. 3K

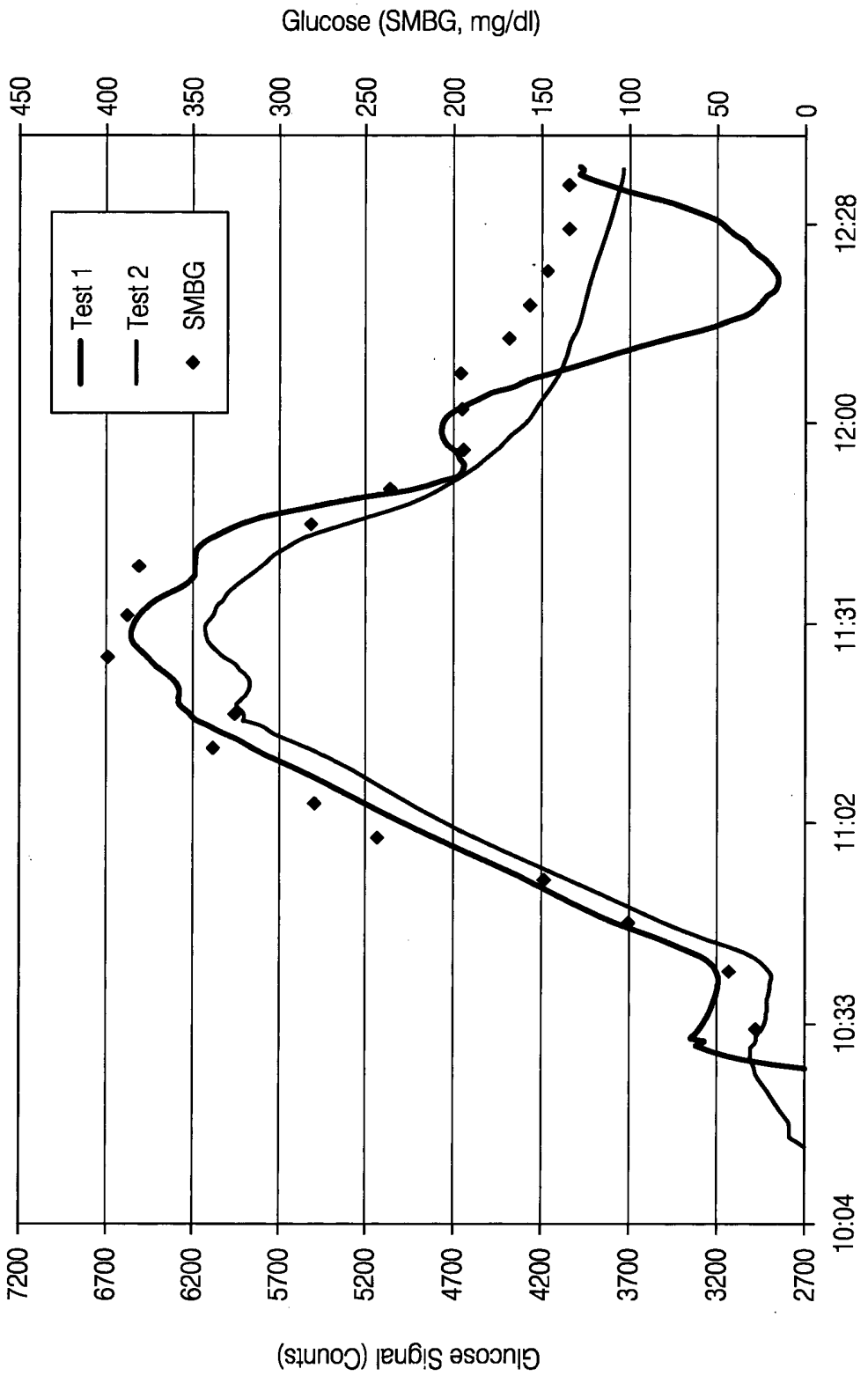
Glucose (SMBG, mg/dl)
450
400
350
300
250
200
150
100
50
0
Test 1
Test 2
SMBG
7200
6700
6200
5700
5200
4700
4200
3700
3200
2700
Glucose Signal (Counts)
10:04
10:33
11:02
11:31
12:00
12:28
Time

FIG. 4

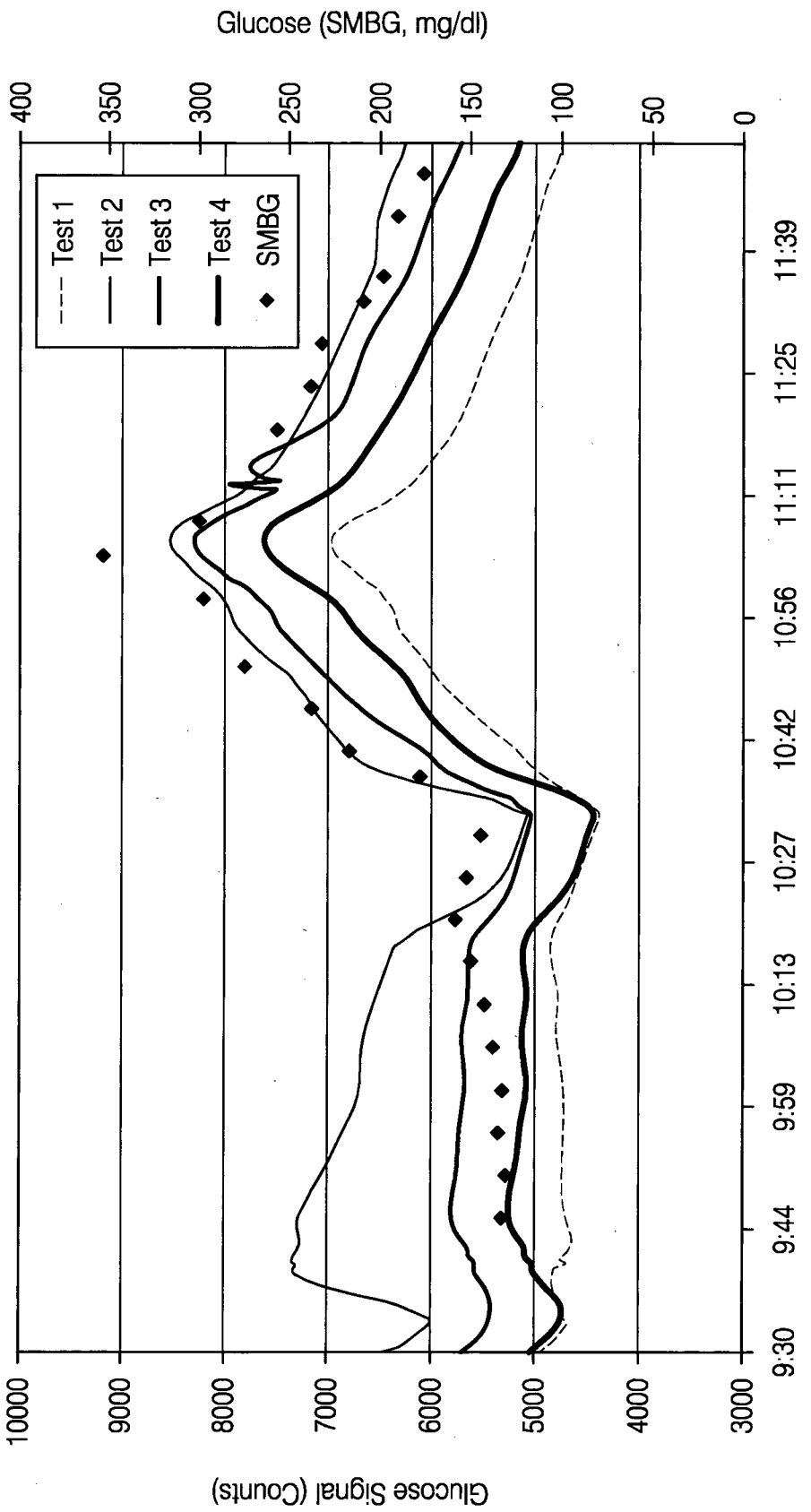
Glucose (SMBG, mg/dl)
400
350
300
250
200
150
100
50
0
Test 1
Test 2
Test 3
Test 4
SMBG
10000
9000
8000
7000
6000
5000
4000
3000
Glucose Signal (Counts)
9:30
9:44
9:59
10:13
10:27
10:42
10:56
11:11
11:25
11:39
Time

FIG. 5

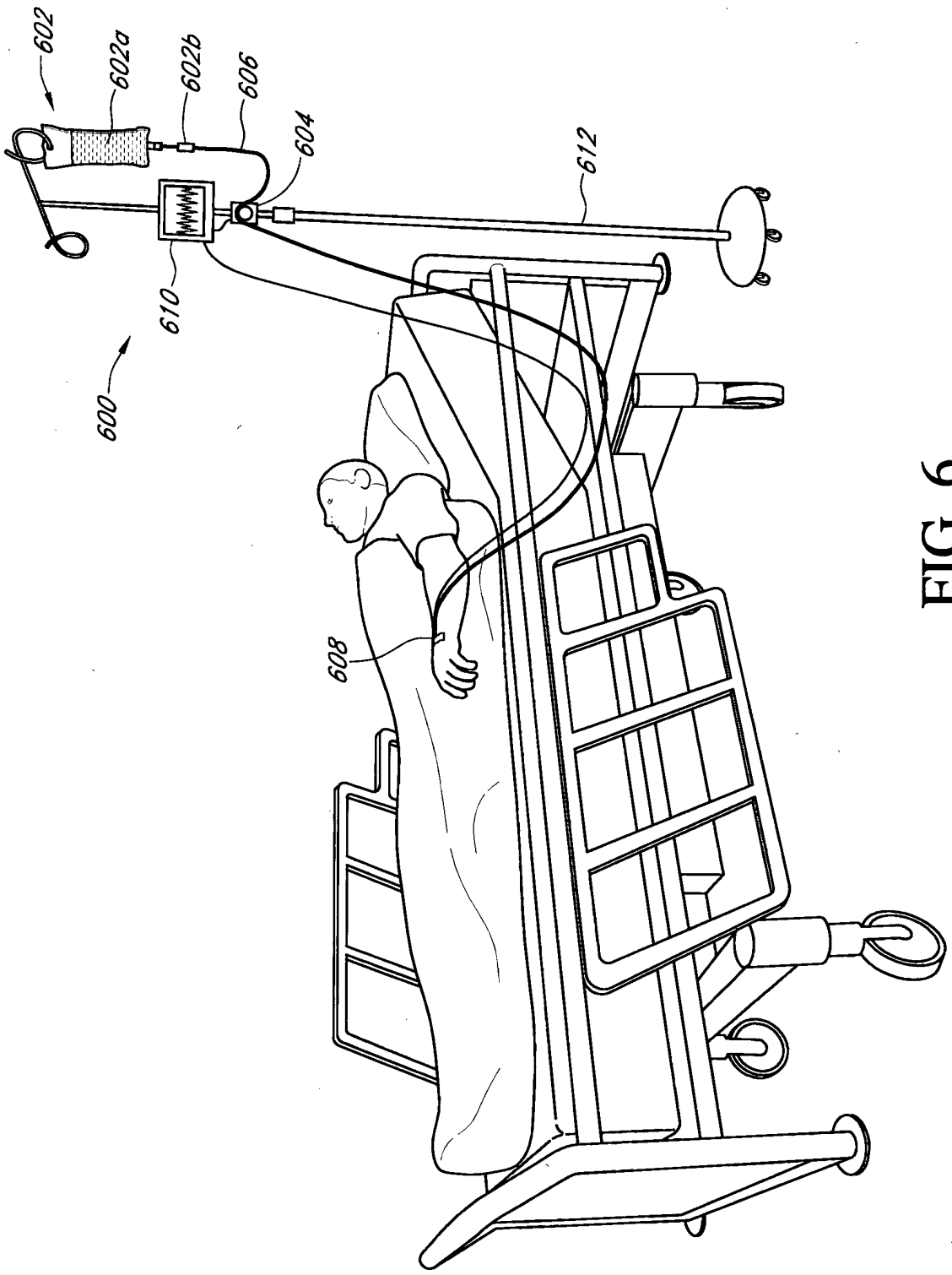


FIG. 6

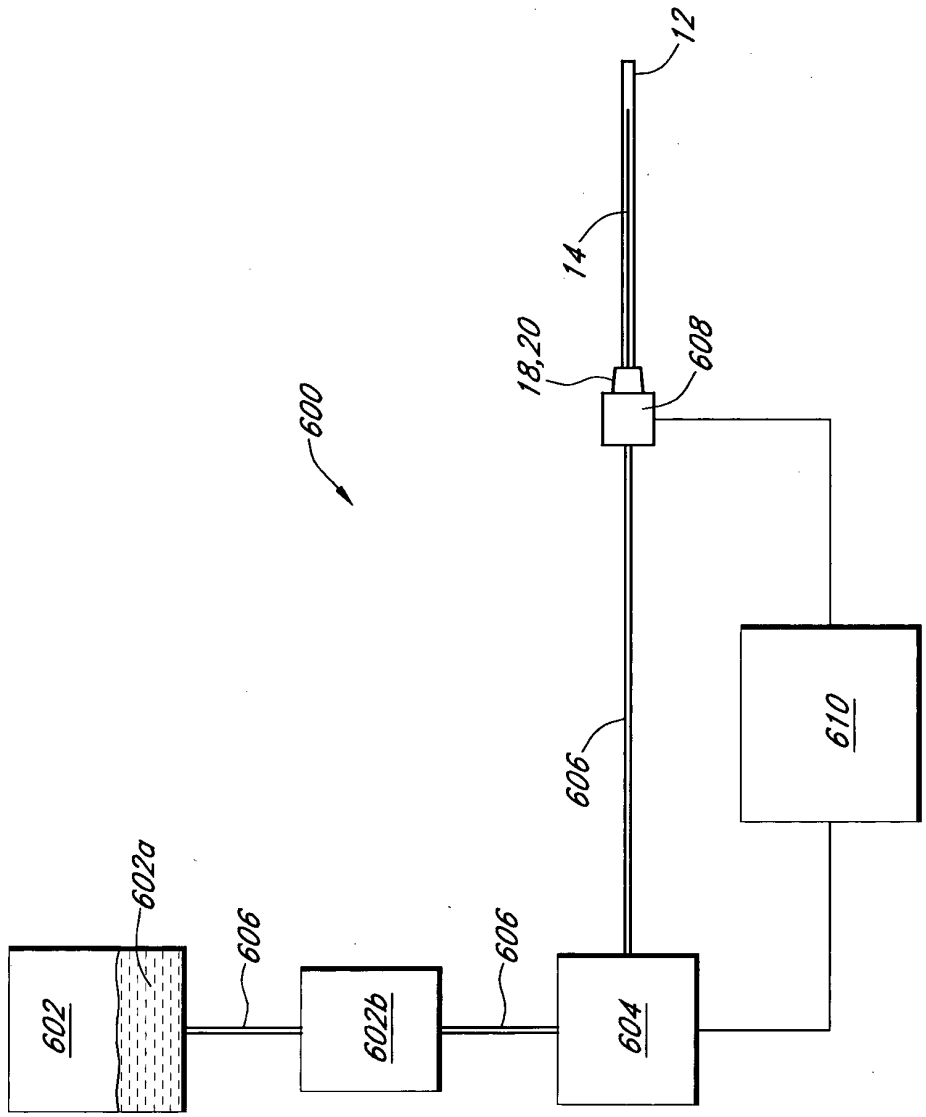


FIG. 7

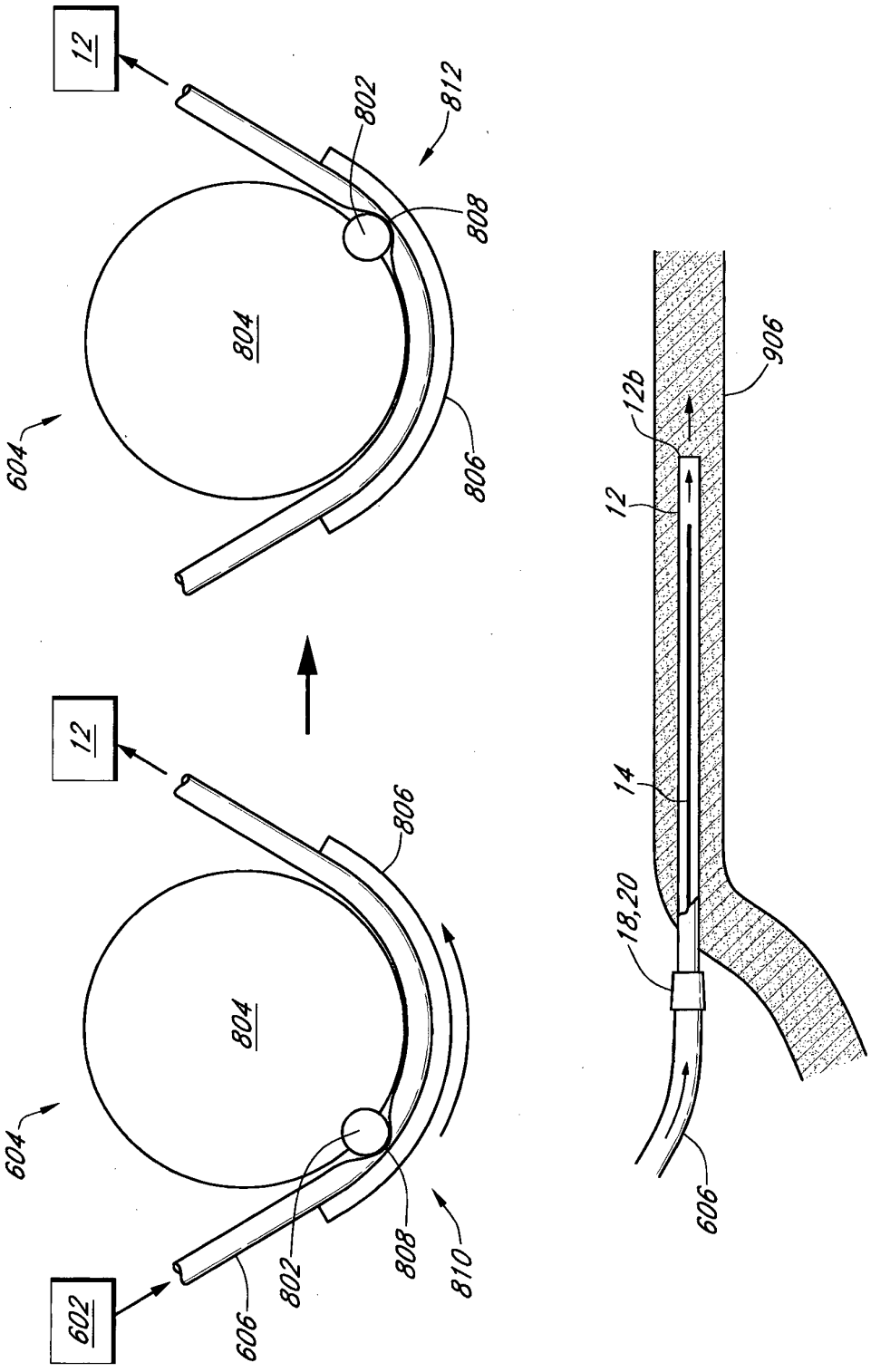
12
802
812
808
804
604
806
12
806
804
604
602
606
802
808
810
906
12b
12
14
18,20
606

FIG. 8A

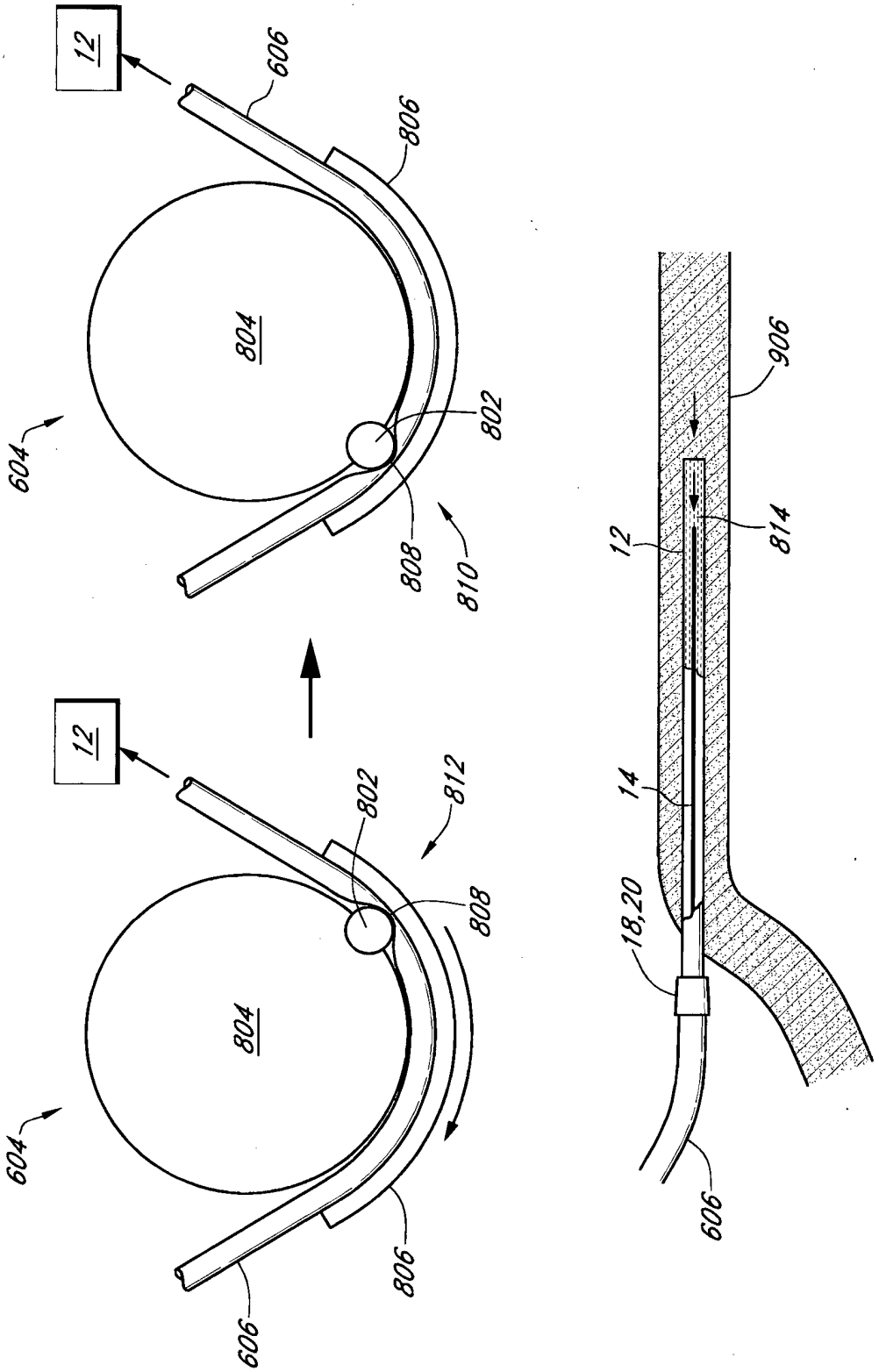


FIG. 8B

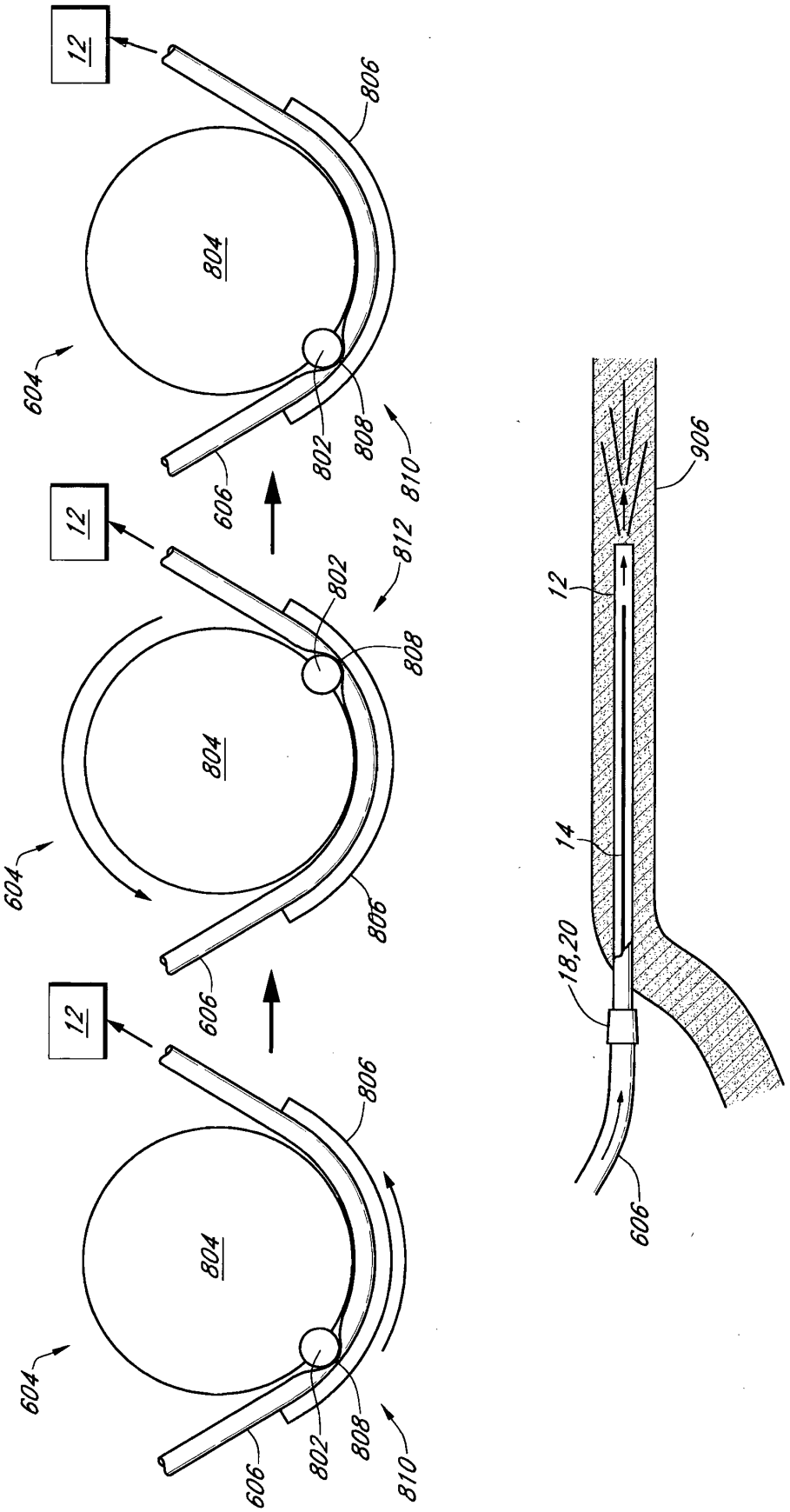


FIG. 8C

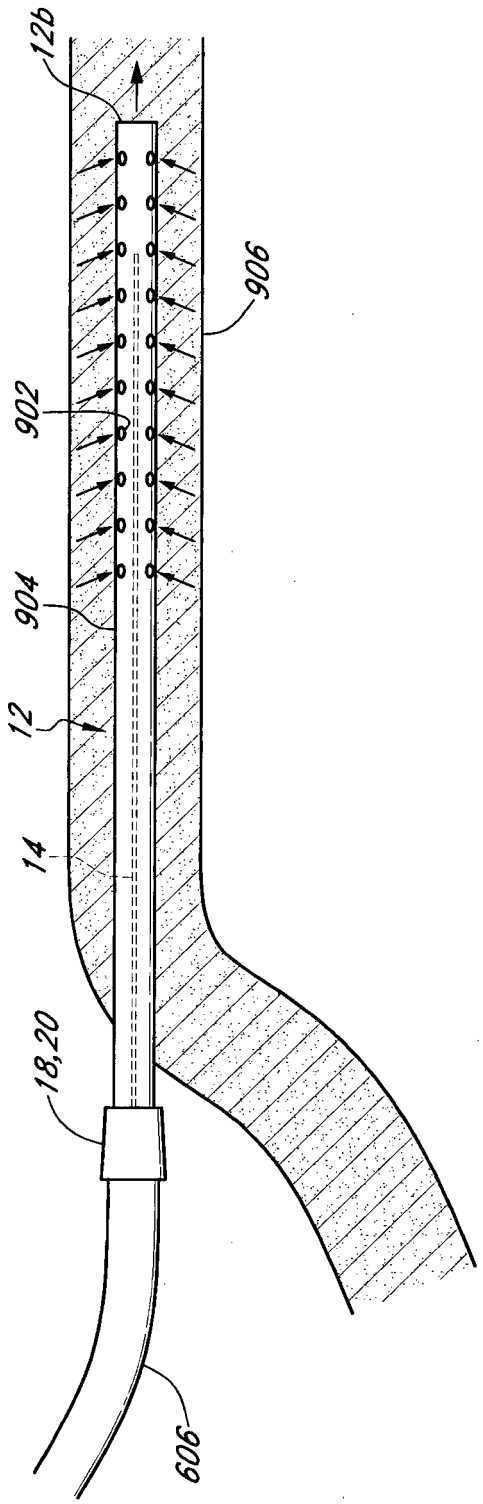


FIG. 9

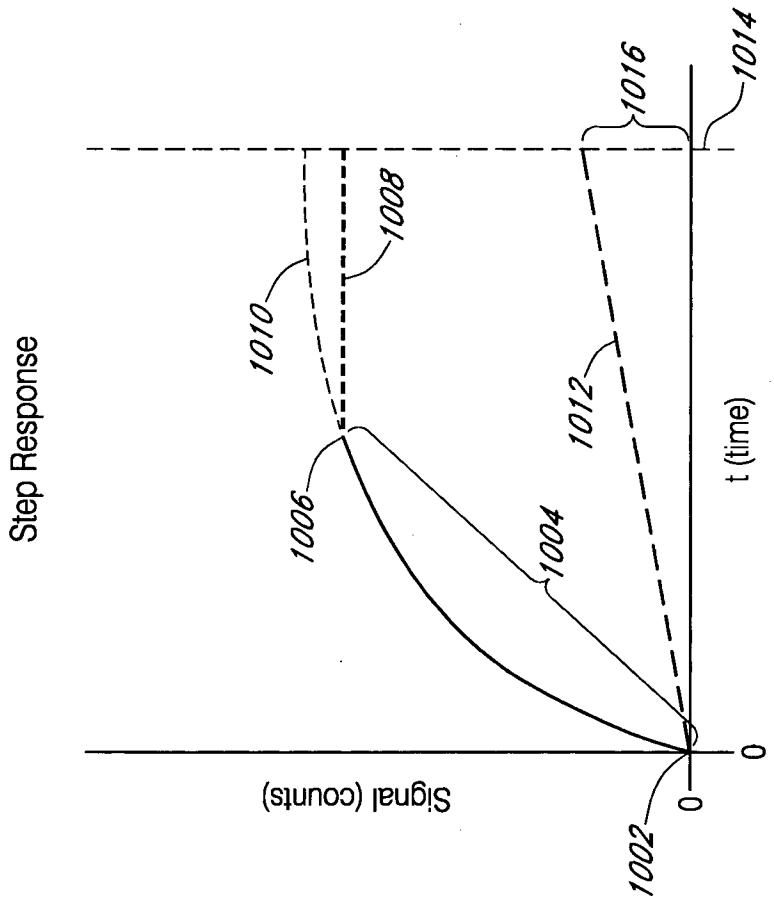
Step Response
Signal (counts)
t (time)
0
0
1002
1004
1006
1008
1010
1012
1014
1016

FIG. 10

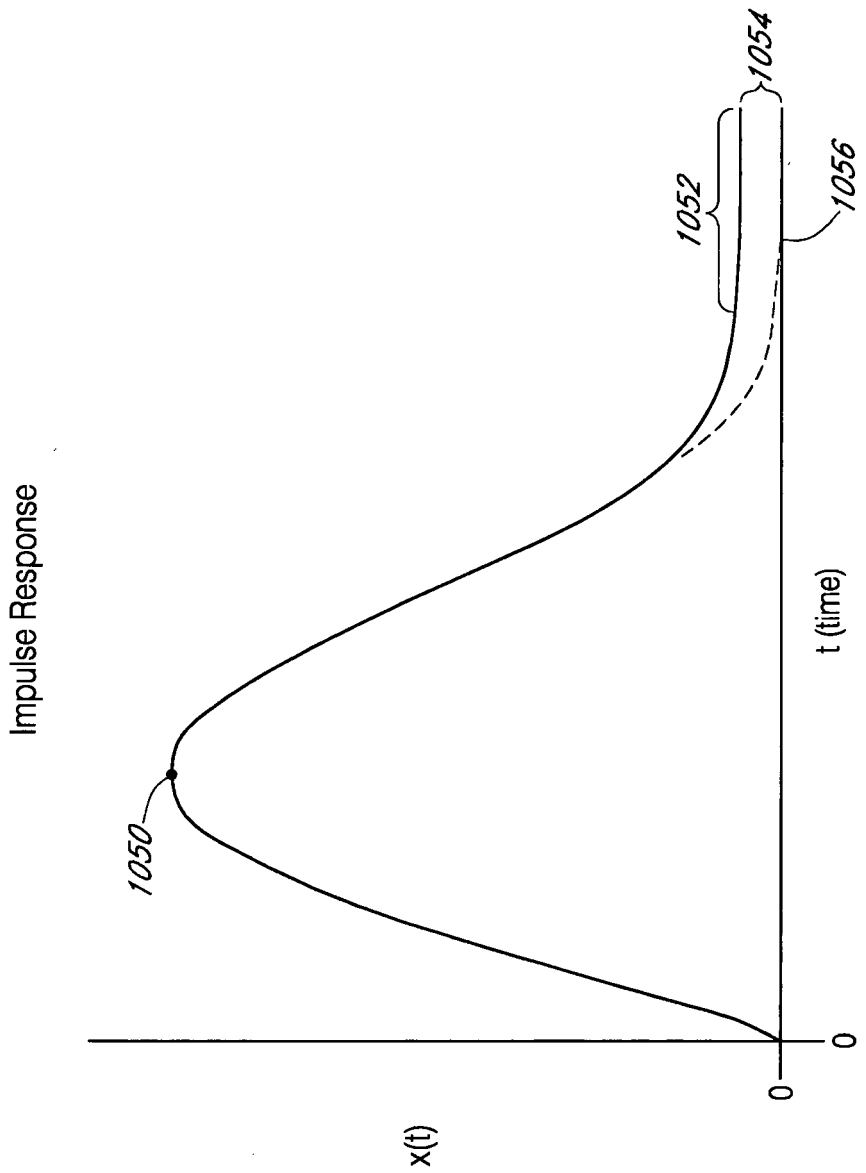


FIG. 11

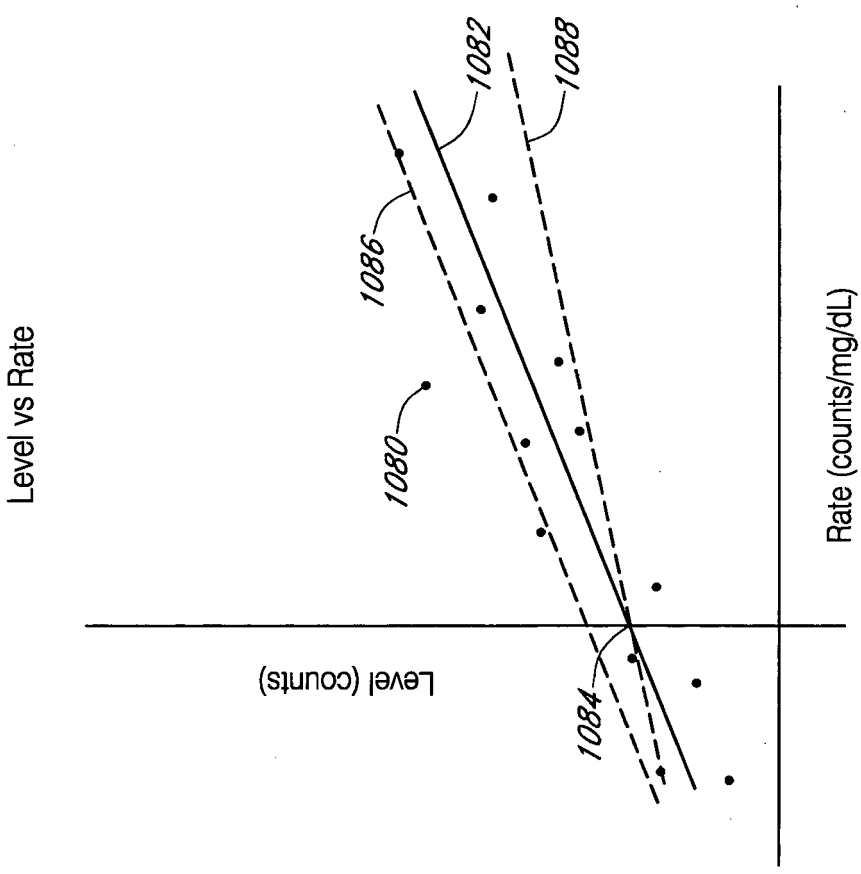
Level vs Rate
Level (counts)
Rate (counts/mg/dL)
1080
1082
1084
1086
1088

FIG. 12

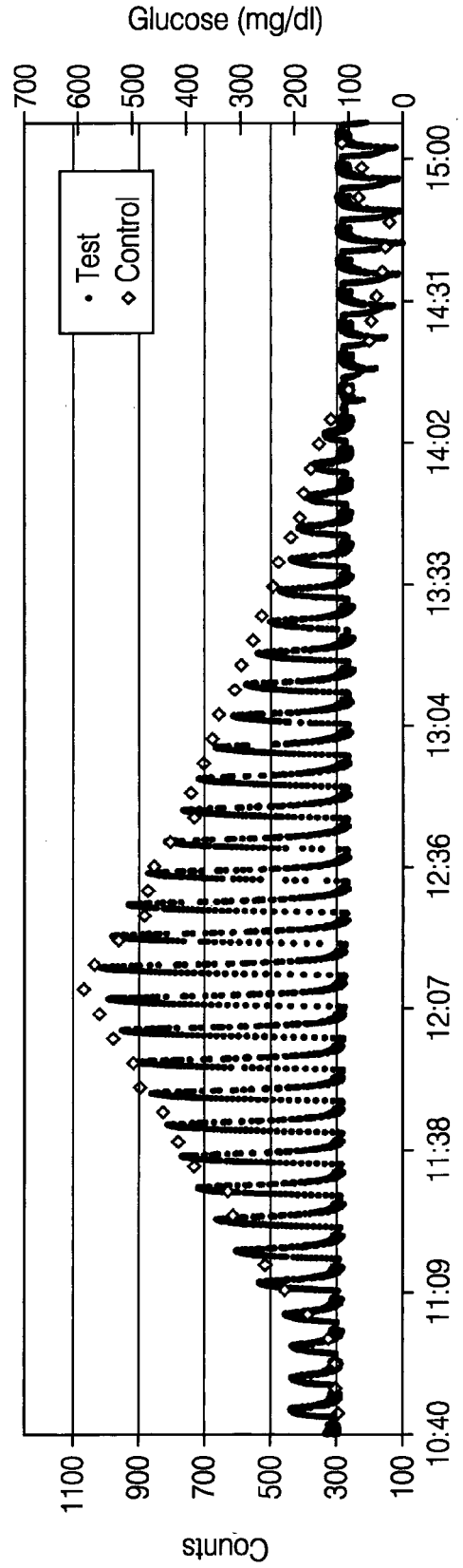
Glucose (mg/dl)
700
600
500
400
300
200
100
0
• Test
◇ Control
15:00
14:31
14:02
13:33
13:04
12:36
12:07
11:38
11:09
10:40
time
1100
900
700
500
300
100
Counts

FIG. 13

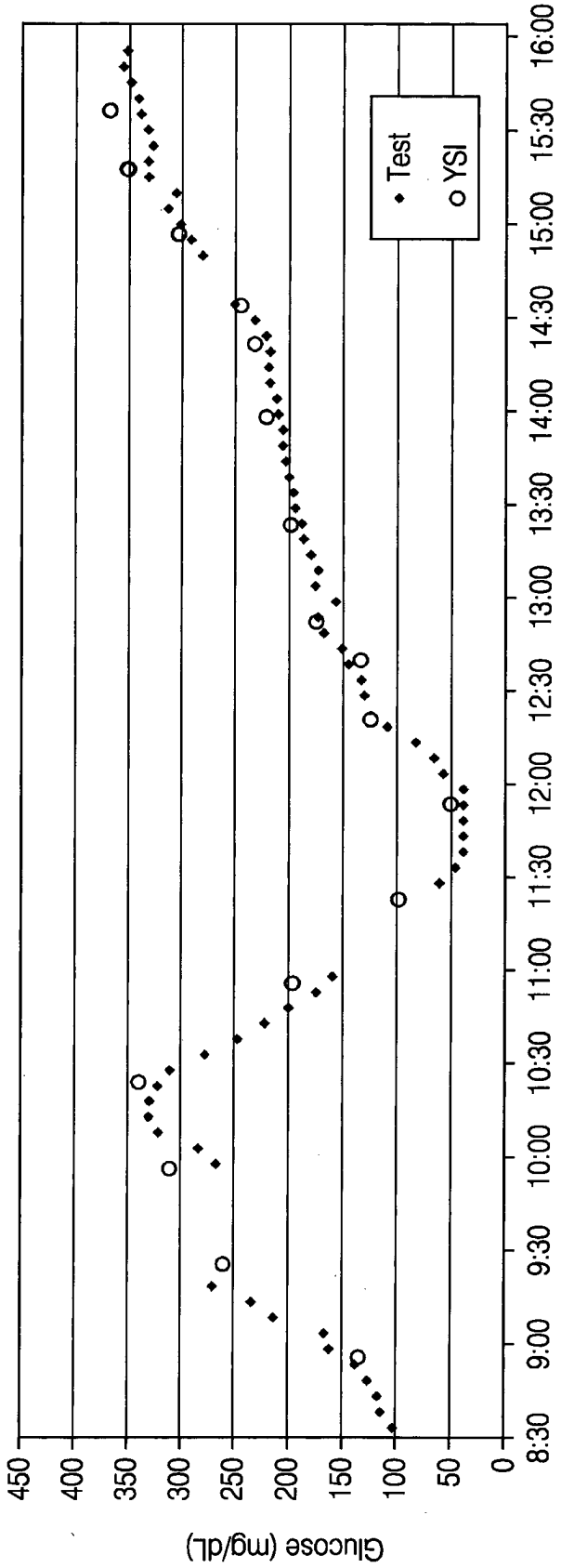
Glucose (mg/dL)
450
400
350
300
250
200
150
100
50
0
8:30
9:00
9:30
10:00
10:30
11:00
11:30
12:00
12:30
13:00
13:30
14:00
14:30
15:00
15:30
16:00
time
Test
YSI

FIG. 14

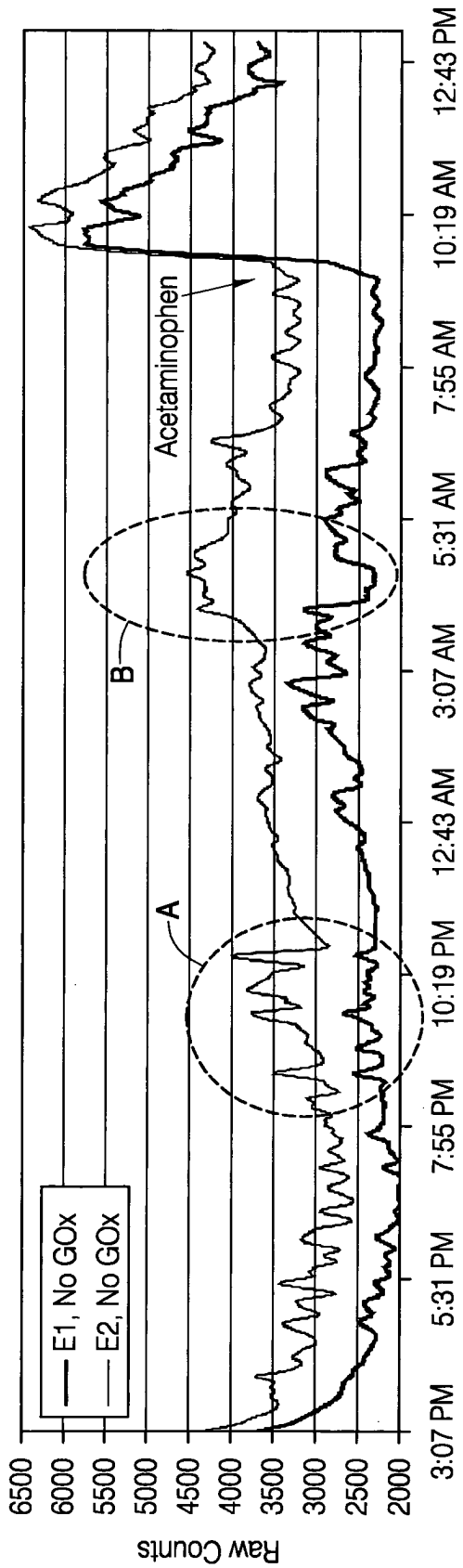
Raw Counts
6500
6000
5500
5000
4500
4000
3500
3000
2500
2000
3:07 PM
5:31 PM
7:55 PM
10:19 PM
12:43 AM
3:07 AM
5:31 AM
7:55 AM
10:19 AM
12:43 PM
E1, No GOx
E2, No GOx
Acetaminophen
A
B

FIG. 15A

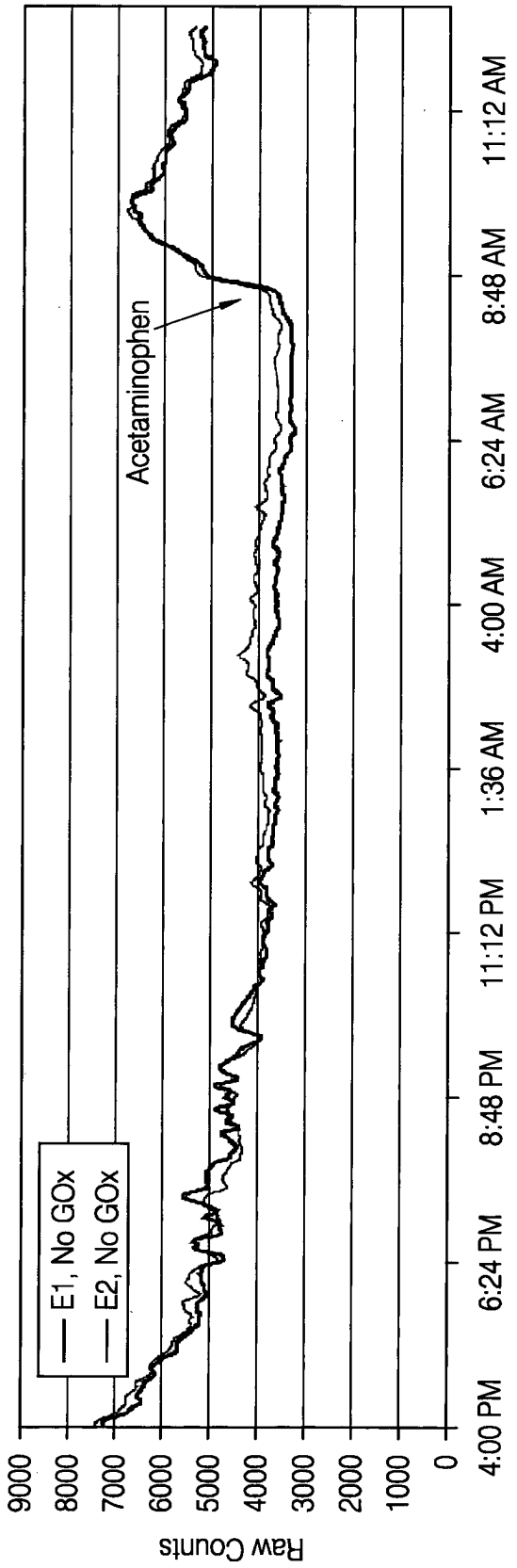
Raw Counts
9000
8000
7000
6000
5000
4000
3000
2000
1000
0
4:00 PM
6:24 PM
8:48 PM
11:12 PM
1:36 AM
4:00 AM
6:24 AM
8:48 AM
11:12 AM
E1, No GOx
E2, No GOx
Acetaminophen

FIG. 15B

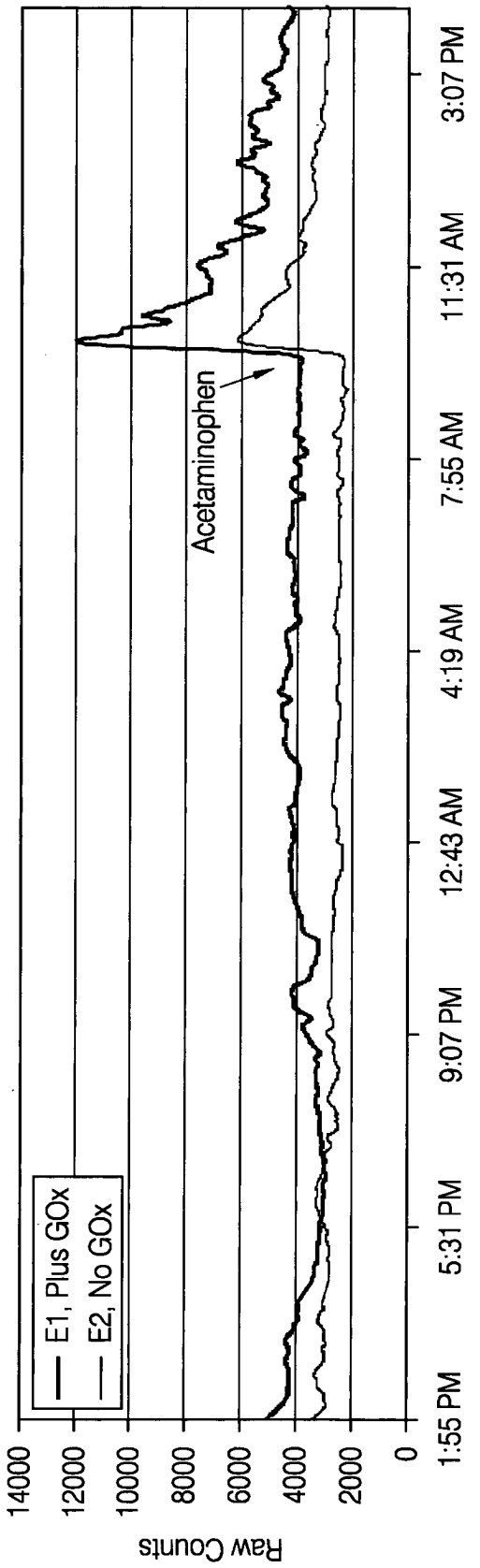
Raw Counts
14000
12000
10000
8000
6000
4000
2000
0
1:55 PM
5:31 PM
9:07 PM
12:43 AM
4:19 AM
7:55 AM
11:31 AM
3:07 PM
E1, Plus GOx
E2, No GOx
Acetaminophen

FIG. 16A

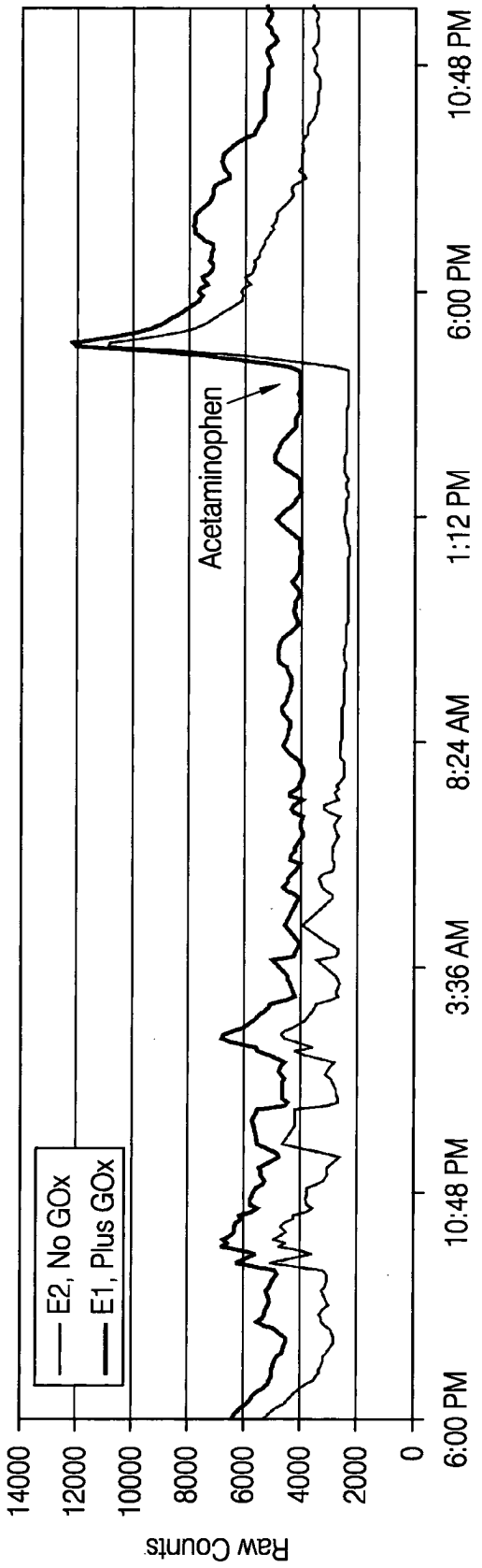
Raw Counts
14000
12000
10000
8000
6000
4000
2000
0
6:00 PM
10:48 PM
3:36 AM
8:24 AM
1:12 PM
6:00 PM
10:48 PM
E2, No GOx
E1, Plus GOx
Acetaminophen

FIG. 16B

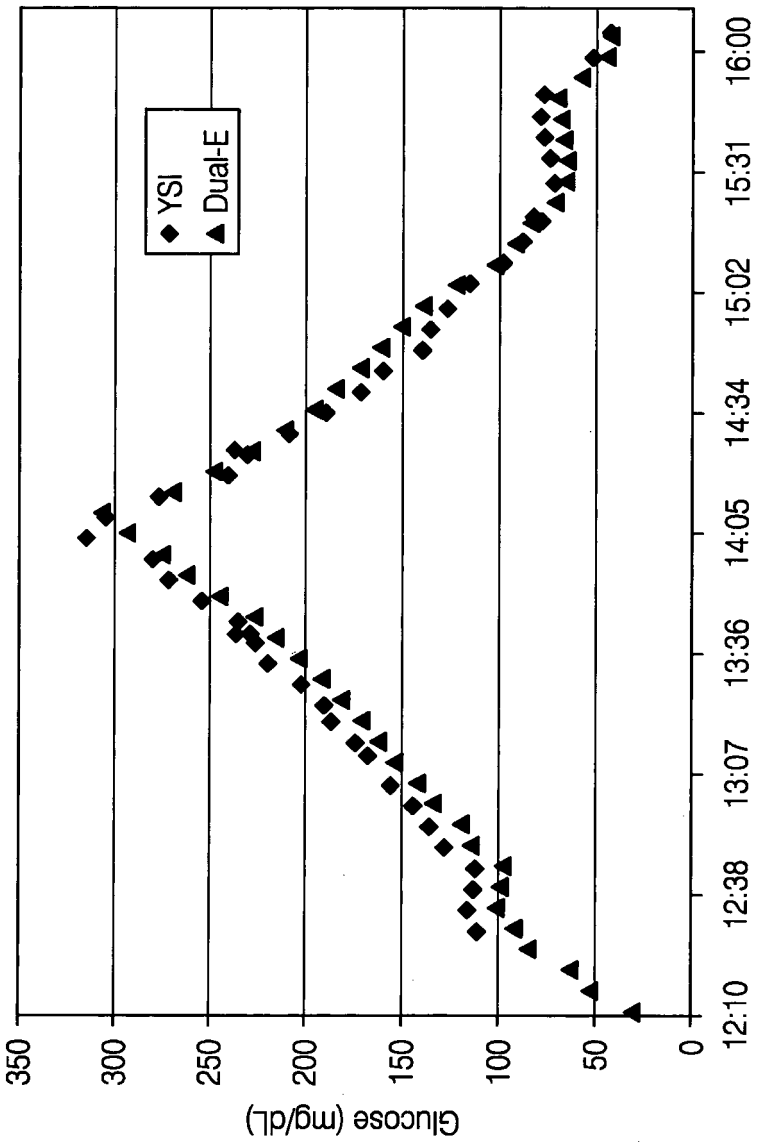
YSI
Dual-E
Glucose (mg/dL)
350
300
250
200
150
100
50
0
12:10
12:38
13:07
13:36
14:05
14:34
15:02
15:31
16:00
Time

FIG. 17A

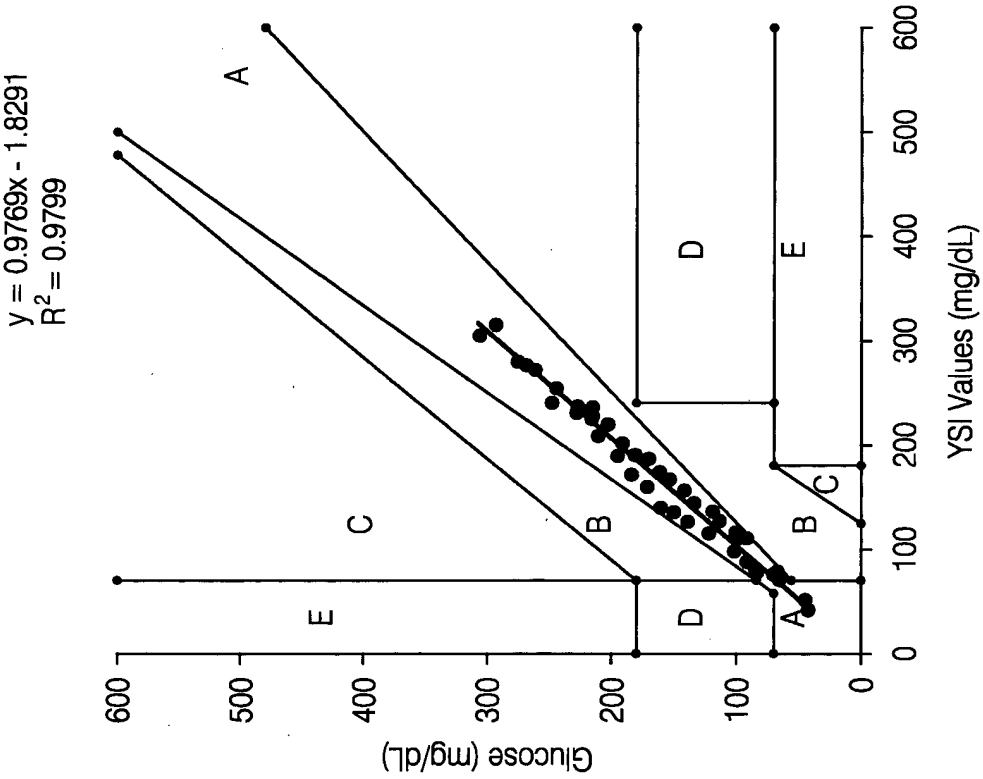
y = 0.9769x - 1.8291
R2 = 0.9799
Glucose (mg/dL)
YSI Values (mg/dL)
0
100
200
300
400
500
600
A
B
C
D
E

FIG. 17B

## REFERENCES CITED IN THE DESCRIPTION

*This list of references cited by the applicant is for the reader's convenience only. It does not form part of the European patent document. Even though great care has been taken in compiling the references, errors or omissions cannot be excluded and the EPO disclaims all liability in this regard.*

### Patent documents cited in the description


- US 6128519 A **[0004]**
- US 4736748 A **[0005]**
- US 4994167 A **[0247] [0917]**
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摘要(译)

提供了用于主机的血管系统的连续分析物测量的系统和方法。 在一些实施例中，连续葡萄糖测量系统包括血管进入装置，传感器和传感器电子设备，该系统被配置用于插入与宿主的循环系统通信。 血管进入装置可包括多个分析物传感器（240），具有管腔（212a）和小孔（212b）的导管（212）以及包括导管（218b）和导管的连接器或集线器（218）。 孔（218c），其构造成通过一个或多个凸缘（218a）与其他静脉内设备连接。