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(54) ACTIVE PULSE SPECTROPHOTOMETRY

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Description

BACKGROUND OF THE INVENTION

[0001] Various spectral photometric techniques have been developed for the noninvasive monitoring of blood constituent concentrations. In such systems, light of multiple wavelengths is used to illuminate a thin tissue portion of a person, such as a fingertip or earlobe. A spectrum analysis of light transmitted through or reflected from the tissue portion is used to measure the light absorption characteristics of blood flowing through the tissue portion. Utilizing calibration data, the concentration of various blood constituents is then derived from known light absorption characteristics of these blood constituents. In one spectral photometric methodology, the absolute optical spectrum of light received from the tissue portion is measured. In a differential spectral photometric methodology such as disclosed in US 5,553,613, blood constituent concentrations are derived from photoplethysmograph data that is responsive to blood volume changes. Pulse oximetry systems, which use the latter methodology to monitor hemoglobin constituents, have been particularly successful in becoming the standard of care for patient oxygen saturation monitoring.

SUMMARY OF THE INVENTION

[0002] Spectraphotometry for the noninvasive monitoring of blood constituents, such as blood glucose and total hemoglobin, to name a few, is highly desirable. For example, current methods for accurately measuring blood glucose involve drawing blood from the subject, which can be onerous for diabetics who must take frequent samples to closely monitor blood glucose levels. Spectraphotometry is described under no scattering conditions by the Beer-Lambert law, which states that the concentration c_i of an absorbant in solution can be determined by the intensity of light transmitted through the solution, knowing the pathlength d_λ, the intensity of the incident light $I_{0,\lambda}$, and the extinction coefficient $\varepsilon_{i,\lambda}$, at a particular wavelength λ. The generalized Beer-Lambert law is expressed as

$$I_\lambda = I_{o,\lambda} e^{-d_\lambda \cdot \mu_{a,\lambda}} \tag{1}$$

$$\mu_{a,\lambda} = \sum_{i=1}^{n} \varepsilon_{i,\lambda} \cdot c_i \tag{2}$$

where $\mu_{a,\lambda}$ is the bulk absorption coefficient and represents the probability of absorption per unit length. Dividing both sides of EQ. **1** by $I_{0,\lambda}$ and taking the logarithm yields

$$\ln\left(I_\lambda / I_{0,\lambda}\right) = -d_\lambda \cdot \mu_{a,\lambda} \tag{3}$$

For n wavelengths, EQS. **2** and **3** can be expressed as

$$\begin{bmatrix} \ln\left(I_{\lambda_1}/I_{0,\lambda_1}\right) \\ \vdots \\ \ln\left(I_{\lambda_n}/I_{0,\lambda_n}\right) \end{bmatrix} = -d \begin{bmatrix} \varepsilon_{1,\lambda_1} & \cdots & \varepsilon_{n,\lambda_1} \\ \vdots & \ddots & \vdots \\ \varepsilon_{1,\lambda_n} & \cdots & \varepsilon_{n,\lambda_n} \end{bmatrix} \begin{bmatrix} c_1 \\ \vdots \\ c_n \end{bmatrix} \tag{4}$$

assuming the pathlength is approximately constant at the wavelengths of interest, i.e. $d_{\lambda_1} = d_{\lambda_2} = \cdots d_{\lambda_n} = d$. EQ. **4** can be rewritten as

$$\mathbf{I}(\lambda) = -d\mathbf{A}(\lambda)\mathbf{C} \tag{5}$$

Solving for the constituent concentrations yields

$$\mathbf{C} = -\frac{1}{d}\mathbf{A}(\lambda)^{-1}\mathbf{I}(\lambda) \tag{6}$$

[0003] As is well known in the art, a system of linear equations can be solved if there are as many linearly independent equations as unknowns. Applied to EQ. **6,** the concentration of a particular blood constituent can be calculated if the number of discrete wavelengths used is equal to the number of significant absorbers that are present and if the absorption characteristics of the significant absorbers are distinguishable at these wavelengths.

[0004] FIG. **1** is a hemoglobin extinction graph **100** illustrating the application of the Beer-Lambert law to pulse oximetry. The extinction graph **100** has an extinction coefficient axis **101** in units of cm^{-1}/mole and a wavelength axis **102** in units of nm. An Hb curve **110** and an HbO_2 curve **160** show the light absorption (extinction) properties of reduced hemoglobin and oxyhemoglobin, respectively. In particular, Hb and HbO_2 have significantly different absorption characteristics in the red to near IR wavelengths. Indeed, Hb absorption **110** is greater than HbO_2 absorption **160** in the red spectrum and, conversely, HbO_2 absorption **160** is greater than Hb absorption **110** in the near IR spectrum. At red and near IR wavelengths below 970 nm, where water has a significant peak, hemoglobin species are the only significant absorbers. Further, Hb and HbO_2 normally are the only hemoglobin species having significant concentrations in blood. Thus, only two wavelengths are needed to resolve the concentrations of Hb and HbO_2. Further, if one red wavelength and one IR wavelength are used, the absorption characteristics of Hb and HbO_2 are different enough at these wavelengths to resolve the concentrations of Hb and HbO_2. Typically, a pulse oximetry sensor utilizes a red emitter, such as a light emitting diode (LED) operating at 660 nm, and an IR emitter, such as a LED operating at 905 nm. As a practical matter, pulse oximetry does not explicitly compute a solution to EQ. **6,** but computes a ratio of concentrations so that the pathlength, d, may be cancelled, as described with respect to EQ. **7,** below.

[0005] FIG. **2** is an absorption chart **200** illustrating the absorption properties of various tissue site components. The absorption chart **200** has a total absorption axis **201** and a time axis **202.** Total absorption **201** is attributed to time-invariant absorption layers **210** and a time-variant absorption layer **260.** The time-invariant absorption layers **210** include a tissue absorption layer **220,** which includes skin, muscle, bone, fat and pigment; a venous blood absorption layer **230;** and a baseline arterial blood absorption layer **240.** The time-variant absorption layer **260** is due to the pulse-added volume of arterial blood, i.e. the differential volume of arterial blood due to the inflow during systole and the outflow during diastole. The time-variant absorption layer **240** has a plethysmograph absorption profile **270.**

[0006] Pulse oximetry relies on the pulsatile nature of arterial blood to differentiate blood constituent absorption from absorption of other constituents in the surrounding tissues. That is, the sensor signal generated by the pulse-added arterial blood layer **260** is isolated from the signal generated by other layers **210,** including tissue, venous blood and baseline arterial blood. To do this, pulse spectraphotometry computes a ratio of the AC portion of the detected signal, which is due to the time-variant layer **260,** with respect to the DC portion of the detected signal, which is due to the time-invariant layers **210,** for each of multiple wavelengths. Computations of AC/DC ratios provide relative absorption measures that compensate for variations in both incident light intensity and background absorption and, hence, are responsive only to the hemoglobin in the arterial blood. As an example, pulse oximetry typically computes a red (RD) AC/DC ratio and an IR AC/DC ratio. Then, a ratio of ratios is computed, i.e.

$$RD/IR = (AC_{RD}/DC_{RD})/(AC_{IR}/DC_{IR}) \tag{7}$$

[0007] The desired oxygen saturation measurement is then computed empirically from this RD/IR ratio.

[0008] In general, pulse spectraphotometry uses multiple wavelength absorption measures, where the number and value of the wavelengths are based on the number of significant absorbers (analytes) and the absorption characteristics of these analytes. Further, pulse spectraphotometry exploits the plethysmograph absorption profile of arterial blood to cancel the time-invariant absorption contributions from other tissue components and normalization to account for variations in incident light at the different wavelengths. In particular, pulse oximetry systems are generally recognized as providing an accurate measurement of blood oxygen through a comparative measurement of oxyhemoglobin and reduced hemoglobin constituents. The application of pulse spectraphotometry to the accurate measurement of other blood con-

stituents, such as glucose concentration or total hemoglobin, however, poses a number of difficulties, as described below.

[0009] FIG. **3** illustrates is an absorption graph **300** for water in the near infrared (IR) spectrum. The absorption graph **300** has an absorption coefficient axis **301** in units of cm^{-1} and a wavelength axis **302** in units of nm. Biological tissues contain a significant percentage of water. Thus, the combination of the light absorption and scattering characteristics of water largely determine the useful range of wavelengths for pulse spectraphotometry. A water absorption curve **310** shows that water absorption increases rapidly with increasing wavelength **302** in the near IR, i.e. in the 750 nm to 3000 nm wavelength range. Fortunately for pulse oximetry, water is not a significant absorber compared with hemoglobin in the red and small wavelength portion of the near infrared, i.e. in the 660 to 940 nm wavelength range. Water, however, is a significant absorber in the larger wavelength portion of the near infrared and beyond. In particular, the penetration depth in water of wavelengths around about 1400 nm is 1 mm or less, and the penetration depth decreases rapidly with increasing wavelength beyond 1400 nm.

[0010] Some of the blood constituents of interest are not significant absorbers in the range of wavelengths where photons can penetrate biological tissue. For example, glucose is not a significant absorber in the visible spectrum. Glucose does have strong absorption bands in the far IR, having an absorption peak at 9700 nm, but photon penetration depth at that wavelength is on the order of 10 μm, i.e. around three orders of magnitude less than in the visible and near IR bands used in pulse oximetry.

[0011] When very little analyte is present, such as for blood glucose, the resulting low signal-to-noise ratio (SNR) represents an inherent system limitation for pulse spectraphotometry in the near IR. Further, there are multiple blood constituents, such as hemoglobin, cholesterol and various proteins, such as albumin and gammaglobulins that are significant absorbers in the near IR. Thus, unlike pulse oximetry, more than two wavelengths are required to resolve a particular analyte at these wavelengths.

[0012] One aspect of a pulse spectraphotometry system is a light source adapted to illuminate a tissue site with optical radiation having a plurality of wavelengths selected from one of a primary band and a secondary band, where the tissue site has a modulated blood volume. A detector is configured to receive the optical radiation attenuated by the tissue site and to generate a detector output responsive to absorption of the optical radiation within the tissue site. A normalizer operating on the detector output generates a plurality of normalized plethysmographs corresponding to the plurality of wavelengths. Further, a processor is configured to calculate a ratio of fractional volumes of analytes in the blood volume based upon the normalized plethysmographs. In a preferred embodiment, the primary band is in a range of 1620 nm to 1730 nm and the secondary band is in a range of 1000 nm to 1380 nm. In a more preferred embodiment, the primary band is in a range of 1620 nm to 1670 nm. In a most preferred embodiment, at least one of the wavelengths is selected in a range of 1650 nm ± 5 nm, in a range of 1032 nm ± 5 nm, in a range of 1097 nm ± 5 nm or in a range of 1375 nm ± 5 nm.

[0013] Another aspect of a pulse spectraphotometry system is a pulse spectraphotometry method comprising the steps of illuminating a tissue site having a pulsatile blood flow with a narrowband optical radiation, time division multiplexing the optical radiation over a plurality of wavelengths and selecting at least a portion of the wavelengths within a range of 1620 nm to 1730 nm. Further steps include detecting an attenuated optical radiation from the tissue site as the result of the illuminating step and calculating a ratio of analytes in the blood flow based upon the attenuated optical radiation. In a preferred embodiment, the selecting step comprises the substep of selecting at least one wavelength in a range of 1620 nm to 1670 nm. In a more preferred embodiment, the selecting step comprises a substep of selecting at least one wavelength in a range of 1650 nm ± 5 nm. In another preferred embodiment, the pulse spectraphotometry method comprises the further step of selecting a second portion of the wavelengths within a range of 1000 nm to 1380 nm. In another more preferred embodiment, the selecting a second portion step comprises a substep of selecting at least one wavelength in a range of 1032 nm ± 5 nm, in a range of 1097 nm ± 5 nm or in a range of 1375 nm ± 5 nm.

[0014] Yet another aspect of a pulse spectraphotometry system is an optical radiation means for illuminating a tissue site and a filter means for determining a nominal wavelength of the optical radiation, where the nominal wavelength is selected from a range of 1620 nm to 1730 nm. The pulse spectraphotometry system also has a detector means for receiving the optical radiation after transmission through or reflection from the tissue site and for generating a corresponding plethysmograph signal and a processor means for calculating a ratio of analyte portions of pulsatile blood flow within the tissue site based upon the plethysmograph signal. In a preferred embodiment, the nominal wavelength is selected from a range of 1620 nm to 1670 nm. In a more preferred embodiment, the nominal wavelength is selected from a range of 1650 nm ± 5 nm. In another preferred embodiment, the nominal wavelength is selected from a range of 1000 nm to 1380 nm. In another more preferred embodiment, the nominal wavelength is selected from a range of 1032 nm ± 5 nm, a range of 1097 nm ± 5 nm, or a range of 1375 nm ± 5 nm. In yet another preferred embodiment, the nominal wavelength is selected from a range of 2000 nm to 2500 nm.

BRIEF DESCRIPTION OF THE DRAWINGS

[0015]

FIG. **1** is a graph of oxyhemoglobin and reduced hemoglobin extinction coefficients versus wavelength;
FIG. **2** is a graph of total tissue and blood light absorption versus time;
FIG. **3** is a graph of water absorption versus wavelength across a portion of the near infrared spectrum;
FIG. **4** is a functional block diagram of a pulse spectraphotometry system;
FIG. **5** is a functional block diagram of a physiological sensor for pulse spectraphotometry; and
FIG. **6** is a graph of water and glucose absorption versus wavelength including critical wavelength portions of the near infrared spectrum for pulse spectraphotometry.

Detailed Description of the Preferred Embodiment

[0016] FIG. **4** illustrates a pulse spectraphotometry system having a signal processor **400** and a sensor **500.** Herein, the term pulse spectraphotometry is intended to include spectraphotometry based upon the pulsatile characteristics of arterial blood, as described above, and spectraphotometry based upon an active pulse, as described below. The sensor **500** is described in detail with respect to FIG. **5,** below. The signal processor **400** has an analog-to-digital converter (ADC) **410,** a demodulator **420,** an active pulse control **430,** a wavelength control **440,** a timing control **450,** a normalizer **470**, a matrix calculator **480** and a ratio calculator **490.** The sensor **500** illuminates a tissue site **10** (FIG. **5)** with multiple wavelengths, one at a time, and generates a signal output **412** that is responsive to the intensity of optical radiation absorbed by the tissue site **10** (FIG. **5).** The signal output **412** is a time division multiplexed (TDM) signal with multiple time slots corresponding to the multiple wavelengths. The timing control **450** determines the time slots. The ADC **410** digitizes the signal output **412,** and the demodulator **420** separates the individual time slots and corresponding responses. The demodulator output **422** is then normalized **470,** such as by dividing the AC by the DC, as described above, to generate a normalized photoplethysmograph (NPP) **472.**

[0017] Taking into account scattering in the tissue media and the resulting wavelength dependent optical pathlengths, the transmitted intensity through the media and, hence, the sensor signal output **412** can be approximated as

$$I_\lambda \approx A_\lambda e^{-mpl_\lambda \cdot \mu_{a,\lambda}} \tag{8}$$

which is similar in form to EQ. **1,** described above, where A_λ is a function of the incident light, the geometry of the tissue media and the tissue media composition; mpl_λ is the wavelength dependent mean pathlength; and $\mu_{a,\lambda}$ is the bulk absorption coefficient expressed in EQ. **2,** above. The NPP **472** can be derived from EQ. **8** as follows

$$dI_\lambda = -mpl_\lambda \cdot A_\lambda e^{-mpl_\lambda \cdot \mu_{a,\lambda}} \cdot d\mu_{a,\lambda} \tag{9}$$

$$NPP = \frac{AC_\lambda}{DC_\lambda} = \frac{dI_\lambda}{I_\lambda} = -mpl_\lambda \cdot d\mu_{a,\lambda} \tag{10}$$

Assume

$$d\mu_{a,\lambda} \approx \mu_{ab_\lambda} \cdot \Delta V / V \tag{11}$$

where μ_{ab_λ} is the bulk absorption coefficient of the blood, V is the tissue volume and ΔV is the change in tissue volume due to the pulsatile blood flow. EQ. **10** can then be written as

$$NPP = -mpl_\lambda \cdot \mu_{ab_\lambda} \cdot v_b \tag{12}$$

$$\mu_{ab_\lambda} = \sum_i \nu_i \cdot \mu_i; \qquad \sum_i \nu_i = 1 \tag{13}$$

which is similar in form to EQS. **2** and **3,** described above, where $\nu_b = \Delta V / V$ is the fractional blood volume. Also, ν_i is the fractional volume in the blood of the ith analyte and μ_i is the absorption coefficient of the *i*th analyte. For n wavelengths, EQS. **12** and **13** can be expressed as

$$\begin{bmatrix} NPP_1 \\ \vdots \\ NPP_n \end{bmatrix} = -\begin{bmatrix} mpl_{\lambda_1} & \cdots & 0 \\ \vdots & \ddots & \vdots \\ 0 & \cdots & mpl_{\lambda_n} \end{bmatrix} \begin{bmatrix} \mu_{1,\lambda_1} & \cdots & \mu_{n,\lambda_1} \\ \vdots & \ddots & \vdots \\ \mu_{1,\lambda_n} & \cdots & \mu_{n,\lambda_n} \end{bmatrix} \begin{bmatrix} \nu_1 \\ \vdots \\ \nu_n \end{bmatrix} \nu_b \tag{14}$$

EQ. **14** can be rewritten as

$$\mathbf{NPP} = -\mathbf{MPL}(\lambda)\boldsymbol{\mu}(\lambda)\mathbf{V}\nu_b \tag{15}$$

Solving for the analyte fractional volumes yields

$$\mathbf{V} = -[\mathbf{MPL}(\lambda)\boldsymbol{\mu}(\lambda)]^{-1}\mathbf{NPP}/\nu_b \tag{16}$$

which is similar in form to EQ. **6.** The matrix calculator **480** performs the matrix inversion indicated in EQ. **16**. The ratio calculator **490** is then used to cancel ν_b- and determine a desired ratio of fractional volumes of analytes in the blood. The mean pathlengths can be determined by calibration or separate measurements. A pathlength measurement method is described in US Patent Application No. 2002/049372. In particular, using the sensor **500** and the wavelengths described in detail with respect to FIG. **6,** below, the ratio calculator output **492** advantageously provides the ratio of the fractional volume of blood glucose to the fractional volume of water in the blood, which is desired for diabetes diagnosis and monitoring.

[0018] Also shown in FIG. **4,** an active pulse control **430** provides an active pulse control input to the sensor **500.** Advantageously, an active pulse mechanism induces a periodic change in the flow of blood through a tissue medium, which can provide a larger AC signal, as described with respect to FIG. **2,** above, and a greater SNR as a result. The degree of blood flow modulation is determined by feedback from the NPP **472** so as to control the AC signal level. The active pulse mechanism is described in further detail with respect to FIG. **5,** below.

[0019] Further shown in FIG. **4,** a wavelength control **440** provides a wavelength control input **442** to the sensor **500.** The wavelength control **440** receives input from the timing control **450** and, accordingly, determines the wavelength sequence of optical radiation illuminating the tissue site **10** (FIG. **5)** in a manner analogous to the LED drive signals transmitted from a pulse oximetry monitor to a pulse oximetry sensor, which is well-known in the art. The sensor wavelength control is described in further detail with respect to FIG. **5,** below.

[0020] FIG. **5** illustrates a pulse spectraphotometry sensor **500** having an ambient light cover **502,** a light source **510,** a condensor lens **520,** a narrow-band, multiple-wavelength filter **530,** a mirror drive **540,** a collecting lens **550,** a fiberoptic cable **560,** a detector **570,** an amplifier **580** and an active pulse transducer **590.** The light source **510** and condensor lens **520** provide broadband optical radiation to the wavelength filter **530,** which passes selected, narrowband portions of the optical radiation to the collecting lens **550.** The narrowband optical radiation is coupled to the fiberoptic cable **560,** which transmits the narrowband optical radiation to illuminate a tissue site **10** that is shielded from ambient light by the cover **502.** The detector **570** generates a current proportional to the intensity of attenuated optical radiation received after transmission through or reflection from the illuminated tissue site **10.** The received intensity is responsive to the absorption coefficients of blood constituents, as described with respect to EQS. **8-13,** above. An amplifier **580** provides a gain in the detector current and generates a signal output **412** to the processor **400** (FIG. **4),** described above.

[0021] As shown in FIG. **5,** the wavelength filter **530** has an input mirror **532,** an output mirror **534,** a parabolic mirror

536 and an optical filter array **538.** The input mirror **532** and output mirror **534** are rotatable according to drive signals **542** from the mirror drive **540.** As such, the mirror drive **540** controls the optical path of light around the parabolic mirror **536** and through a particular optical filter in the optical filter array **538.** Each optical filter in the optical filter array **538** is manufactured to a different narrow passband. Thus, the wavelength filter **530** determines the nominal wavelength of optical radiation that illuminates the tissue site **10** at any particular time. The wavelength control input **442** from the signal processor **400** (FIG. **4)** synchronizes the timing of the input mirror **532** and output mirror **534** rotations and, accordingly, the tissue illumination wavelength and the characteristics of the TDM signal output **412,** described with respect to FIG. **4,** above.

[0022] Also shown in FIG. **5** is the active pulse transducer **590,** which modulates the blood flow at the tissue site **10** according to the active pulse control input **432,** described with respect to FIG. **4,** above. In one embodiment, the active pulse transducer is a pressure device applied to a patient's digit. The pressure device may be, for example, a cuff having a bladder that periodically fills and empties with a gas or liquid, such as air or water. Although shown separate from the sensor in FIG. **5,** the transducer **590** may be an integral part of the sensor. Active pulse apparatuses and methods are described in US Patent No. 6,151,516 entitled "Active Pulse Blood Constituent Monitoring," incorporated by reference herein.

[0023] In one sensor embodiment, the light source **510** is a high intensity incandescent lamp such that several mw of power is introduced into a tissue site, e.g. a finger, at each wavelength. The filter array **538** utilizes multiple Fabry-Perot optical interference filters each having a 10 nm bandwidth. The detector is a InGaAs photodiode having a 2mm - 3mm diameter, a useful response bandwidth in the range of 850 nm to 1700 nm and the highest possible intrinsic shunt resistance. The amplifier is a transimpedance amplifier such as Analog Devices **743,** having a feedback resistance in the range of 20 to 40 MΩ. The TDM signal output **412** (FIG. **4)** switches wavelengths at a 40 Hz rate.

[0024] FIG. **6** is a graph **600** having a logarithmic absorption axis **601** in units of cm^{-1} versus a linear wavelength axis **602** in units of nm. Plotted on the graph **600** is a water absorption curve **610** and glucose absorption curve **660.** In a preferred embodiment, the pulse spectraphotometry system is adapted to operate in a primary wavelength band **630** having a range of 1620 nm to 1730 nm. In a more preferred embodiment, the pulse spectraphotometry system is adapted to operate in a sub-band of the primary band **630** having a range of 1620 nm to 1670 nm. In a most preferred embodiment, the pulse spectraphotometry system is adapted to operate at a nominal wavelength of 1650 nm $\pm$ 5 nm.

[0025] In a preferred embodiment, the pulse spectraphotometry system may also be adapted to operate in a secondary wavelength band **650** having a range of 1000 nm to 1380 nm. In a more preferred embodiment, the pulse spectraphotometry system may also adapted to operate at a nominal wavelength around 1032 nm $\pm$ 5 nm and/or 1097 nm $\pm$ 5 nm, where water and glucose are isobestic. The pulse spectraphotometry system may also be adapted to operate at a nominal wavelength around 1375 nm $\pm$ 5 nm where water has about the same absorption as in the primary wavelength band. Although the wavelength bands of 1620 nm to 1730 nm and of 1000 nm to about 1380 nm are denoted above as primary and secondary wavelength bands, respectively, the pulse spectraphotometry system may operate solely within the primary wavelength band or solely within the secondary wavelength band.

[0026] As shown in FIG. **6,** the preferred embodiment encompasses a critical range of wavelengths for a pulse spectraphotometry system, as described herein. Water absorption rapidly increases an order of magnitude between 1300 nm and 1400 nm and two orders of magnitude between 1300 nm and 1900 nm. Water accounts for a significant percentage of blood, interstitial fluids and other tissue. Further, the intensity of optical radiation transmitted through tissue decreases exponentially with absorption, as described with respect to EQ. **1,** above. Accordingly, for the same input intensity, the detected output intensity, i.e. the DC output of the detector **570** (FIG. **5),** is roughly 200 times less at 1400 nm as compared to 1300 nm. The result is that the signal drops below the electronic noise present in the photodiode and preamp at wavelengths much above 1300 nm. There is a small range of wavelengths in the primary band **630,** which encompasses the preferred range of wavelengths, where it is very difficult, but advantageous, to operate the pulse spectraphotometry system. Within this range, water absorption drops to a value roughly equal to its value around 1380 nm, and it is possible to obtain a working plethysmograph, but only by minimizing all noise sources, including the elimination of ambient light; the selection of low noise electronic components, such as described with respect to FIG. **5,** above; and careful component layout and interconnection to avoid noise sources such as crosstalk and ground noise, as is well-known in the art.

[0027] There are several advantages of the primary band **630** of wavelengths for the spectraphotometric determination of certain blood constituents. At least five blood constituents of significance are in the near IR, including water, glucose, hemoglobin, urea and protein. In one embodiment, the pulse spectraphotometry system operates over at least five wavelengths for resolution of these analytes. The absorption characteristics of these analytes must be sufficiently different at the operating wavelengths to insure a robust solution, as determined by the condition number of the resulting matrix, as is well known in the art. Operating within the primary band **630** and the secondary band **650** increases the variation in absorption characteristics. In particular, there is a crossover of water absorption and glucose absorption between 1380 nm and 1620 nm, which allows glucose to be more easily distinguished from water.

[0028] Another significant advantage of the primary band **630** is that glucose absorption is an order of magnitude

larger within that band than for the secondary band **650,** i.e. at wavelengths below 1380 nm. Because intensity varies exponentially with absorption, as described with respect to EQ. **1,** measurements derived from the primary band **630** are significantly more sensitive to variations in blood glucose than those derived from the secondary band **650.**

[0029] Yet another advantage to the primary band **630** is that, due to the higher absorption of at least some blood constituents in the primary band, including water and glucose, the plethysmograph signal is larger in magnitude in the primary band **630** than for the secondary band **650.** Although the SNR of the plethysmograph signal is lower in the primary band **630,** the larger absolute magnitude provides a greater dynamic range for blood constituent measurements.

[0030] Although a primary band **630** and secondary band **650** are described above with respect to a sensor **500** (FIG. **5)** configured to transmit light through a tissue site **10** (FIG. **5),** in another embodiment, the sensor **500** (FIG. **5)** may be configured to detect illumination reflected from a tissue site **10** (FIG. **5).** Because such a reflectance sensor suffers less tissue absorption than a transmission sensor, the primary band **630** may be somewhat broader, in the range of about 1575 nm to 1775 nm and a tertiary band of wavelengths in the range of about 2000 nm to about 2500 nm may be utilized.

Claims

1. A pulse spectraphotometry system comprising:

a light source adapted to illuminate a tissue site with optical radiation having a plurality of wavelengths selected from a band, said tissue site having a modulated blood volume;
a detector configured to receive said optical radiation attenuated by said tissue site and to generate a detector output responsive to absorption of said optical radiation within said tissue site;
a normalizer operating on said detector output to generate a plurality of normalized plethysmographic data corresponding to said plurality of wavelengths; and
a processor configured to calculate a ratio of fractional volumes of analytes in said blood volume based upon said normalized plethysmographic data, **characterised by**
the band being in a range of 1620 nm to 1730 nm.

2. The pulse spectraphotometry system according to claim 1 wherein said band is in a range of 1620 nm to 1670 nm.

3. The pulse spectraphotometry system according to claim 2 wherein at least one of said wavelengths is selected in a range of 1650 nm ± 5 nm.

4. A pulse spectraphotometry system comprising:

a light source adapted to illuminate a tissue site with optical radiation having a plurality of wavelengths selected from a band, said tissue site having a modulated blood volume;
a detector configured to receive said optical radiation attenuated by said tissue site and to generate a detector output responsive to absorption of said optical radiation within said tissue site;
a normalizer operating on said detector output to generate a plurality of normalized plethysmographic data corresponding to said plurality of wavelengths; and
a processor configured to calculate a ratio of fractional volumes of analytes in said blood volume based upon said normalized plethysmographic data,

characterised by
the band being in a range of 1000 nm to 1380 nm.

5. The pulse spectraphotometry system according to claim 4 wherein at least one of said wavelengths is selected in a range of 1032 nm ± 5 nm.

6. The pulse spectraphotometry system according to claim 4 wherein at least one of said wavelengths is selected in a range of 1097 nm ± 5 nm.

7. The pulse spectraphotometry system according to claim 4 wherein at least one of said wavelengths is selected in a range of 1375 nm ± 5 nm.

Patentansprüche

1. Pulsspektralphotometriesystem mit:

einer Lichtquelle, die zum Beleuchten eines Gewebebereichs mit optischer Strahlung mit mehreren aus einem Band ausgewählten Wellenlängen geeignet ist, wobei der Gewebebereich ein moduliertes Blutvolumen aufweist;
einem Detektor, der zum Empfangen der vom Gewebebereich gedämpften optischen Strahlung und zum Erzeugen einer Detektorausgabe, die auf die Absorption der optischen Strahlung im Gewebebereich anspricht, konfiguriert ist;
einem Normalisator, der auf die Detektorausgabe wirkt, um mehrere normalisierte plethysmographische Daten zu erzeugen, die den mehreren Wellenlängen entsprechen; und
einem Prozessor, der zum Berechnen eines Verhältnisses von fraktionierten Volumina von Analyten in dem Blutvolumen auf der Basis der normalisierten plethysmographischen Daten konfiguriert ist, **dadurch gekennzeichnet**
dass das Band in einem Bereich von 1620 nm bis 1730 nm liegt.

2. Pulsspektralphotometriesystem nach Anspruch 1, wobei das Band im Bereich von 1620 nm bis 1670 nm liegt.

3. Pulsspektralphotometriesystem nach Anspruch 2, wobei mindestens eine der Wellenlängen aus einem Bereich von 1650 nm ± 5 nm ausgewählt ist.

4. Pulsspektralphotometriesystem mit:

einer Lichtquelle, die zum Beleuchten eines Gewebebereichs mit optischer Strahlung mit mehreren aus einem Band ausgewählten Wellenlängen geeignet ist, wobei der Gewebebereich ein moduliertes Blutvolumen aufweist;
einem Detektor, der zum Empfangen der vom Gewebebereich gedämpften optischen Strahlung und zum Erzeugen einer Detektorausgabe, die auf die Absorption der optischen Strahlung im Gewebebereich anspricht, konfiguriert ist;
einem Normalisator, der auf die Detektorausgabe wirkt, um mehrere normalisierte plethysmographische Daten zu erzeugen, die den mehreren Wellenlängen entsprechen; und
einem Prozessor, der zum Berechnen eines Verhältnisses von fraktionierten Volumina von Analyten in dem Blutvolumen auf der Basis der normalisierten plethysmographischen Daten konfiguriert ist, **dadurch gekennzeichnet**
dass das Band in einem Bereich von 1000 nm bis 1380 nm liegt.

5. Pulsspektralphotometriesystem nach Anspruch 4, wobei mindestens eine der Wellenlängen aus einem Bereich von 1032 nm ± 5 nm ausgewählt ist.

6. Pulsspektralphotometriesystem nach Anspruch 4, wobei mindestens eine der Wellenlängen aus einem Bereich von 1097 nm ± 5 nm ausgewählt ist.

7. Pulsspektralphotometriesystem nach Anspruch 4, wobei mindestens eine der Wellenlängen aus einem Bereich von 1375 nm ± 5 nm ausgewählt ist.

Revendications

1. Système de spectrophotométrie à impulsions comprenant :

une source de lumière adaptée pour éclairer un site de tissu avec un rayonnement optique ayant une pluralité de longueurs d'onde sélectionnées à partir d'une bande, ledit site de tissu ayant un volume de sang modulé ;
un détecteur configuré pour recevoir ledit rayonnement optique atténué par ledit site de tissu et pour générer un résultat de détecteur sensible à l'absorption dudit rayonnement optique à l'intérieur du site de tissu ;
un normalisateur ayant une action sur ledit résultat du détecteur pour générer une pluralité de données pléthysmographiques normalisées correspondant à ladite pluralité de longueurs d'onde ; et
un processeur configuré pour calculer un rapport de volume infime de substances à analyser dans ledit volume

de sang en fonction desdites données pléthysmographiques normalisées, **caractérisé par** la bande qui est de l'ordre de 1 620 nm à 1 730 nm.

2. Système de spectrophotométrie à impulsions selon la revendication 1, dans lequel ladite bande est de l'ordre de 1 620 nm à 1 670 nm.

3. Système de spectrophotométrie à impulsions selon la revendication 2, dans lequel au moins l'une desdites longueurs d'onde est choisie dans une plage de 1 650 nm ± 5 nm.

4. Système de spectrophotométrie à impulsions comprenant :

une source de lumière adaptée pour éclairer un site de tissu avec le rayonnement optique ayant une pluralité de longueurs d'onde choisies à partir d'une bande, ledit site de tissu ayant un volume de sang modulé ;
un détecteur configuré pour recevoir ledit rayonnement optique atténué par ledit site de tissu et pour générer un résultat de détecteur sensible à l'absorption dudit rayonnement optique à l'intérieur dudit site de tissu ;
un normalisateur ayant une action sur ledit résultat du détecteur afin de générer une pluralité de données pléthysmographiques normalisées correspondant à ladite pluralité de longueurs d'onde ; et
un processeur configuré pour calculer un rapport de volume infime de substances à analyser dans ledit volume de sang en fonction desdites données pléthysmographiques normalisées, **caractérisé par** la bande secondaire qui est de l'ordre d'environ 1000 nm à 1 380 nm.

5. Système de spectrophotométrie à impulsions selon la revendication 4, dans lequel au moins l'une desdites longueurs d'onde est choisie dans une plage de 1 032 nm ± .5 nm.

6. Système de spectrophotométrie à impulsions selon la revendication 4, dans lequel au moins l'une desdites longueurs d'onde est choisie dans une plage de 1 097 nm ± 5 nm.

7. Système de spectrophotométrie à impulsions selon la revendication 4, dans lequel au moins l'une desdites longueurs d'onde est choisie dans une plage de 1 375 nm ± 5 nm.

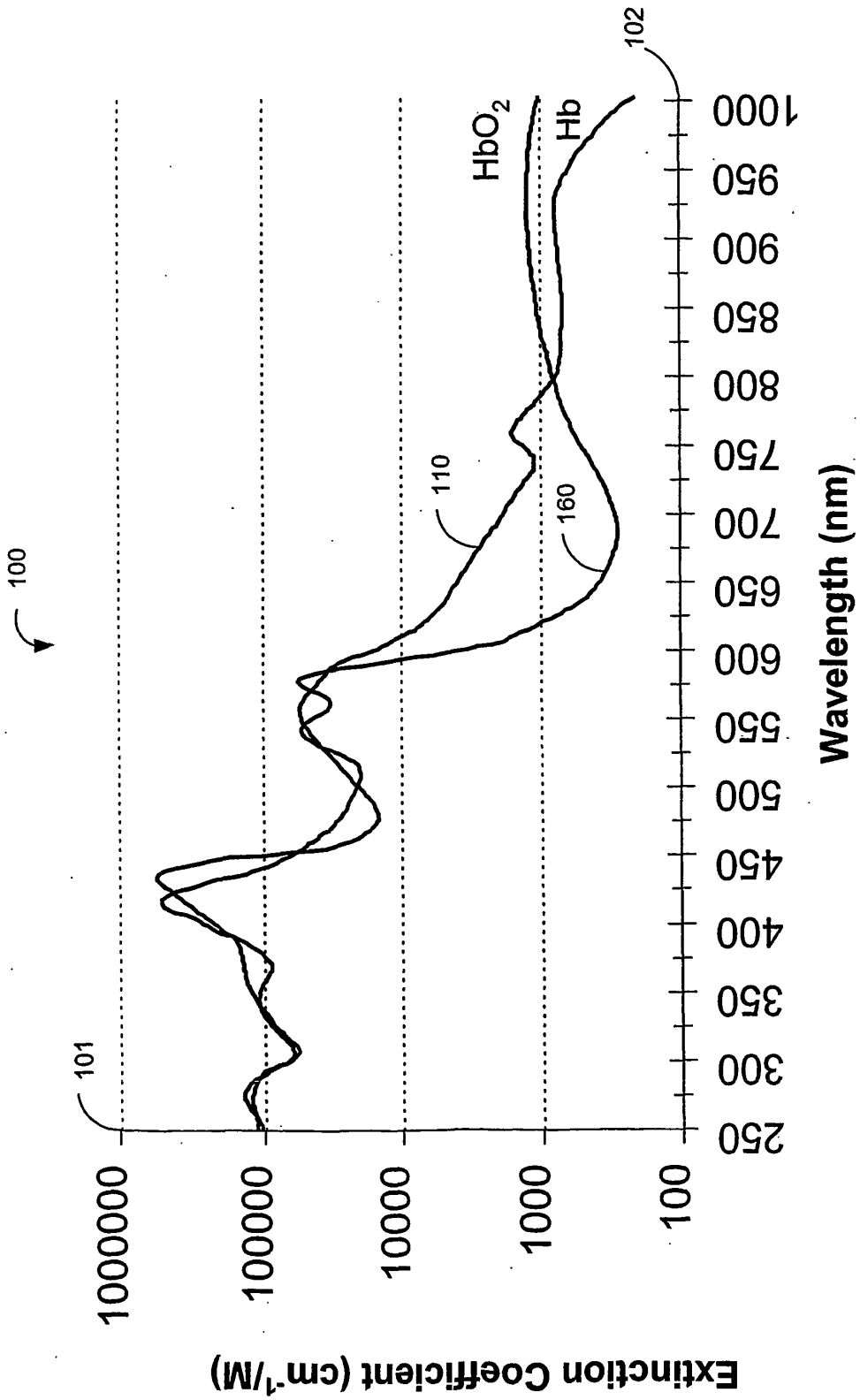
100
101
102
110
160
HbO2
Hb
1000000
100000
10000
1000
100
250
300
350
400
450
500
550
600
650
700
750
800
850
900
950
1000
Extinction Coefficient (cm-1/M)
Wavelength (nm)

FIG. 1

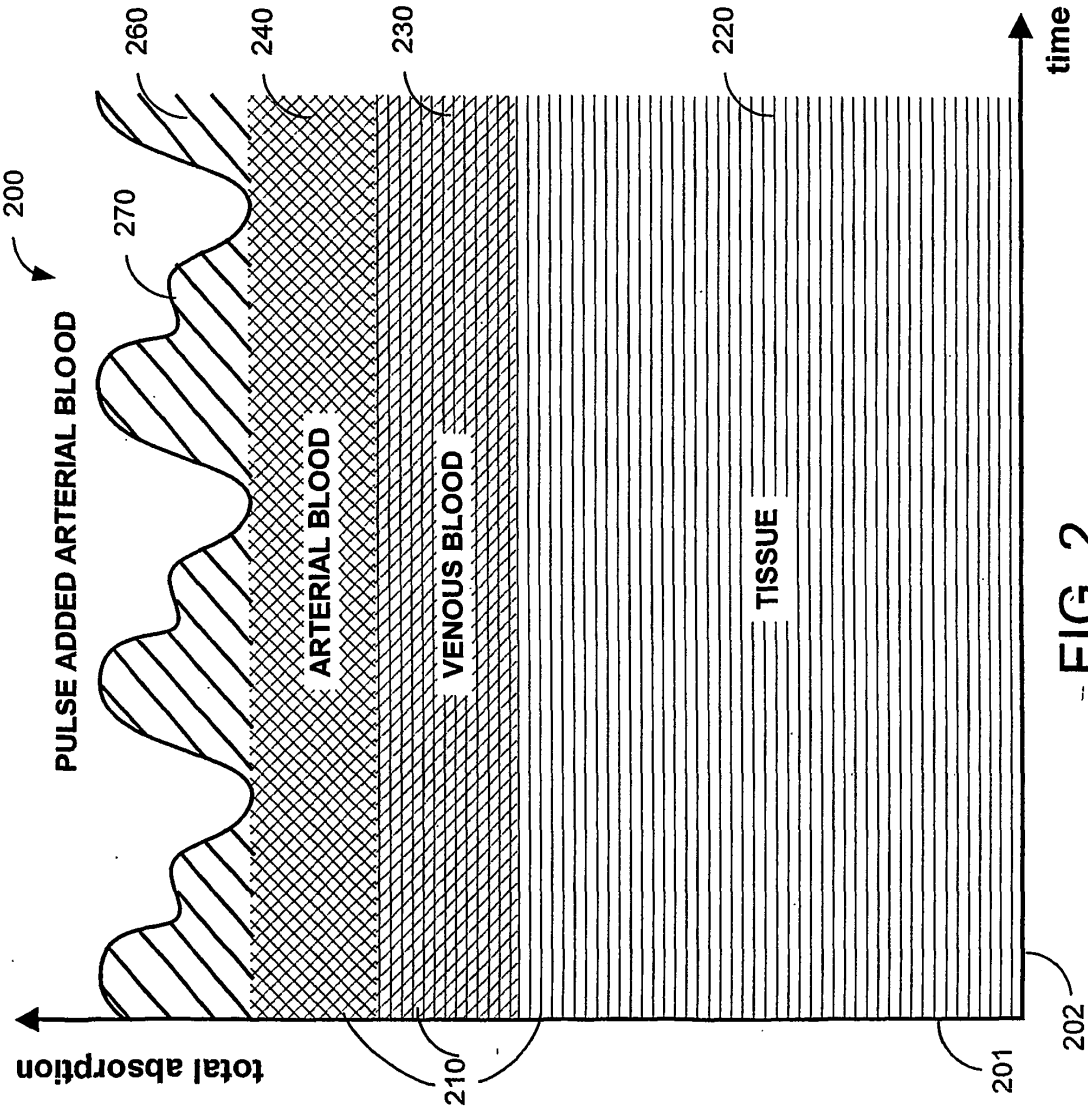
260
240
230
220
time
200
270
PULSE ADDED ARTERIAL BLOOD
ARTERIAL BLOOD
VENOUS BLOOD
TISSUE
total absorption
210
201
202

FIG. 2

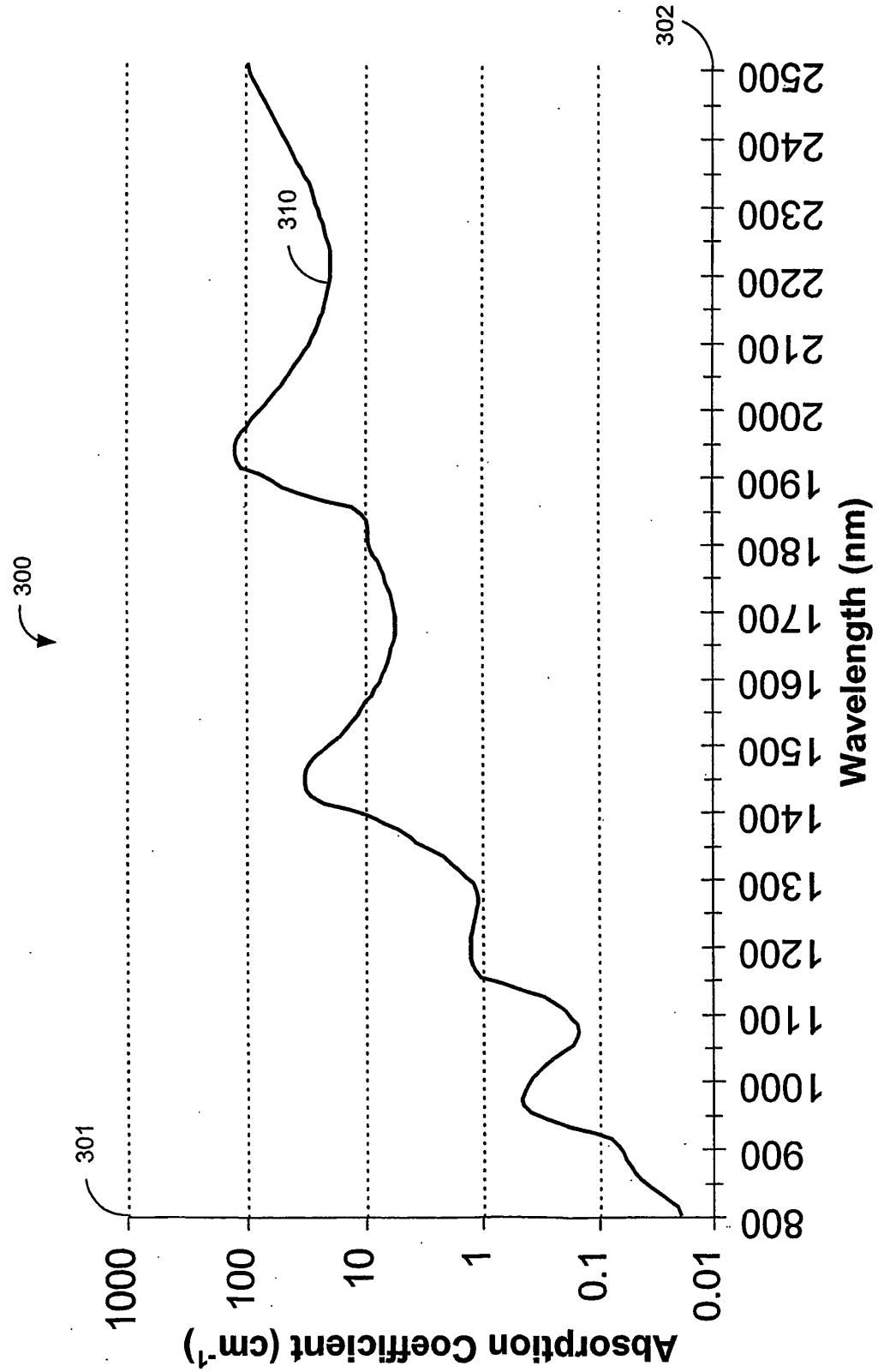
300
301
302
310
Absorption Coefficient (cm-1)
1000
100
10
1
0.1
0.01
Wavelength (nm)
800
900
1000
1100
1200
1300
1400
1500
1600
1700
1800
1900
2000
2100
2200
2300
2400
2500

FIG. 3

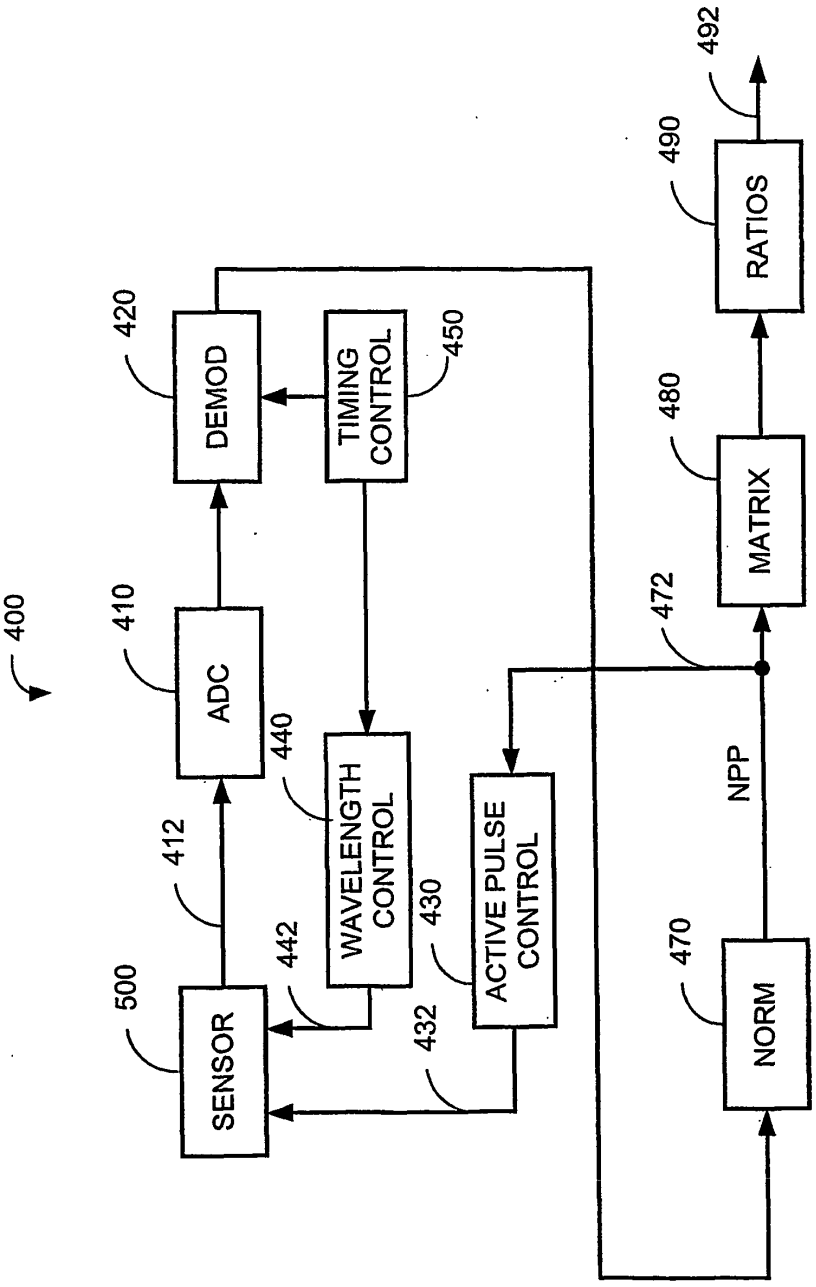
400
410
ADC
412
420
DEMOD
450
TIMING CONTROL
440
WAVELENGTH CONTROL
442
430
ACTIVE PULSE CONTROL
432
500
SENSOR
470
NORM
NPP
472
480
MATRIX
490
RATIOS
492

FIG. 4

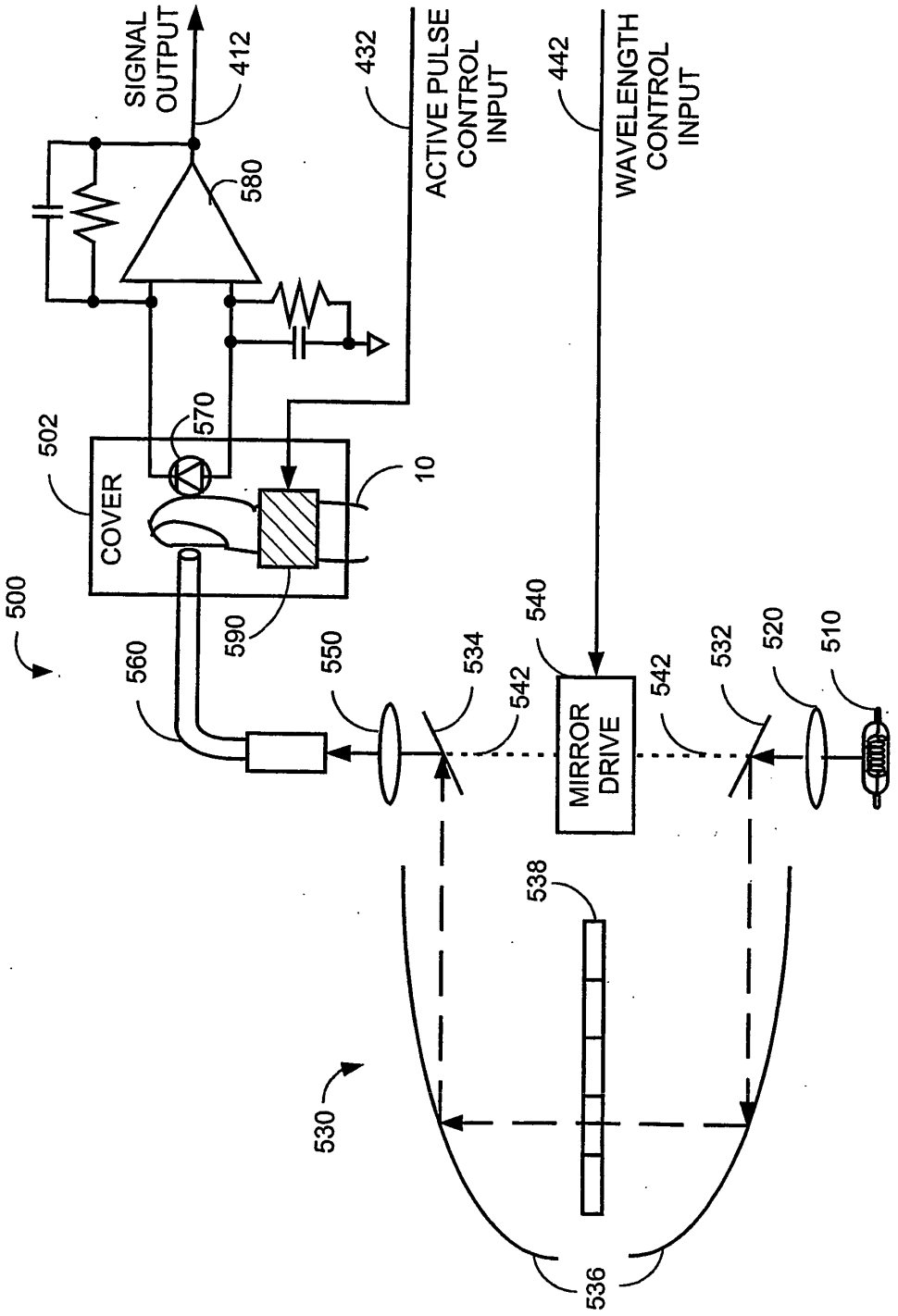
SIGNAL OUTPUT
412
432
ACTIVE PULSE CONTROL INPUT
442
WAVELENGTH CONTROL INPUT
580
570
502
COVER
10
500
590
560
550
534
540
542
MIRROR DRIVE
542
532
520
510
538
530
536

FIG. 5

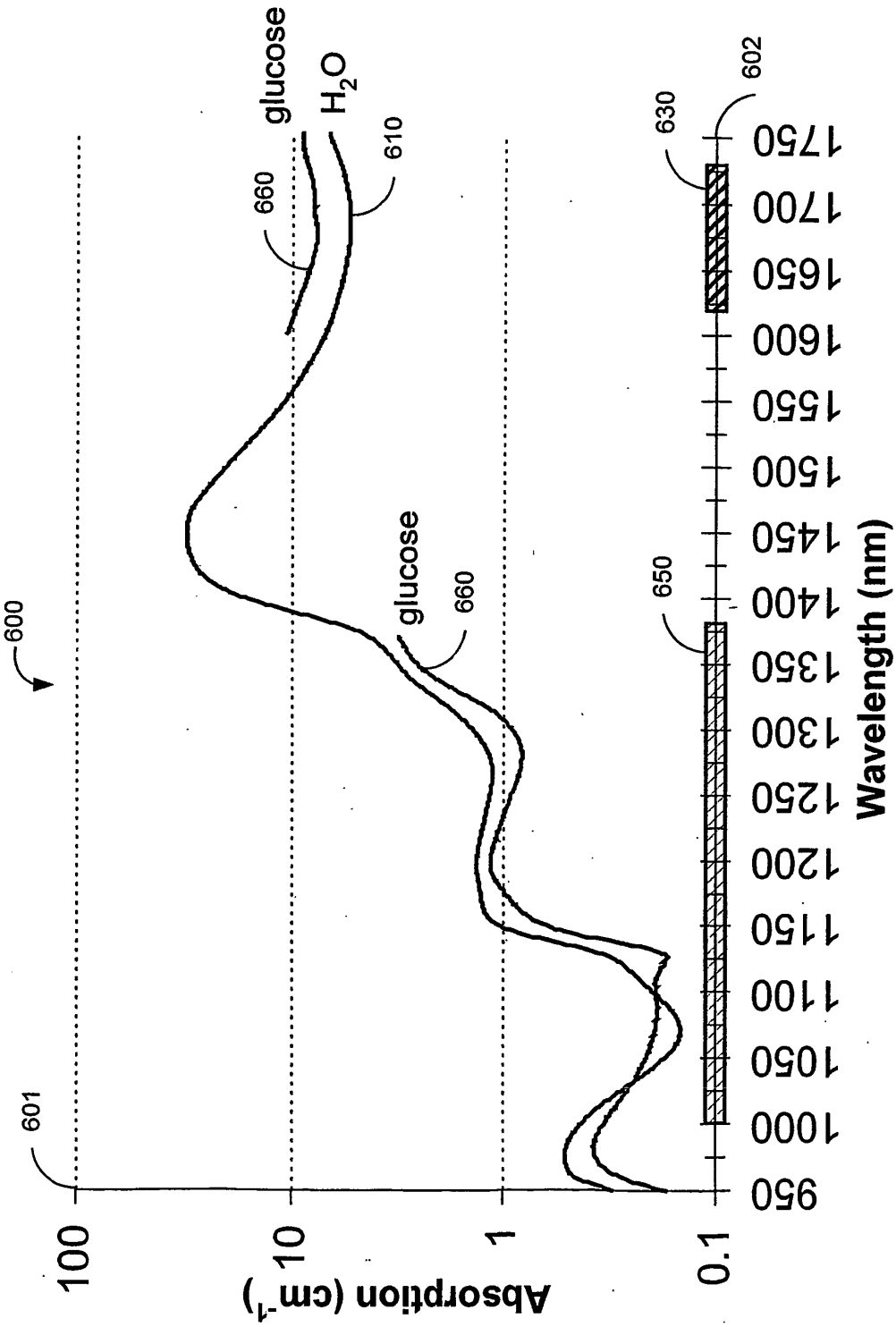
600
601
602
610
630
650
660
660
glucose
glucose
H_2O
100
10
1
0.1
Absorption (cm^{-1})
950
1000
1050
1100
1150
1200
1250
1300
1350
1400
1450
1500
1550
1600
1650
1700
1750
Wavelength (nm)

FIG. 6

REFERENCES CITED IN THE DESCRIPTION

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Patent documents cited in the description

- US 5553613 A **[0001]**
- US 2002049372 A **[0017]**
- US 6151516 A **[0022]**

专利名称(译)	有源脉冲分光光度法		
公开(公告)号	EP1478265B1	公开(公告)日	2010-04-14
申请号	EP2003713552	申请日	2003-02-21
[标]申请(专利权)人(译)	梅西莫股份有限公司		
申请(专利权)人(译)	Masimo公司		
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[标]发明人	DIAB MOHAMED		
发明人	DIAB, MOHAMED		
IPC分类号	A61B5/00		
CPC分类号	A61B5/14532 A61B5/0059 A61B5/1455 A61B5/14552 A61B5/6826 A61B5/6838		
代理机构(译)	法思博事务所		
优先权	60/358809 2002-02-22 US		
其他公开文献	EP1478265A1		
外部链接	Espacenet		

摘要(译)

脉冲和有源脉冲分光光度测定系统包括适于用光辐射照射组织部位的光源，所述光辐射具有从主带和次带中的至少一个中选择的多个波长。组织部位具有由动脉血的脉动性质或诱导脉冲产生的调节血液体积。检测器配置成接收由组织部位衰减的光学辐射，并响应于组织部位内的光学辐射的吸收而产生检测器输出。在检测器输出上操作的标准化器产生对应于多个波长的多个标准化体积描记器。此外，处理器被配置为基于标准化体积描记器计算血液体积中分数体积的分数的比率。