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(54) PULSE OXIMETRY SIGNAL CORRECTION USING NEAR INFRARED ABSORPTION BY WATER

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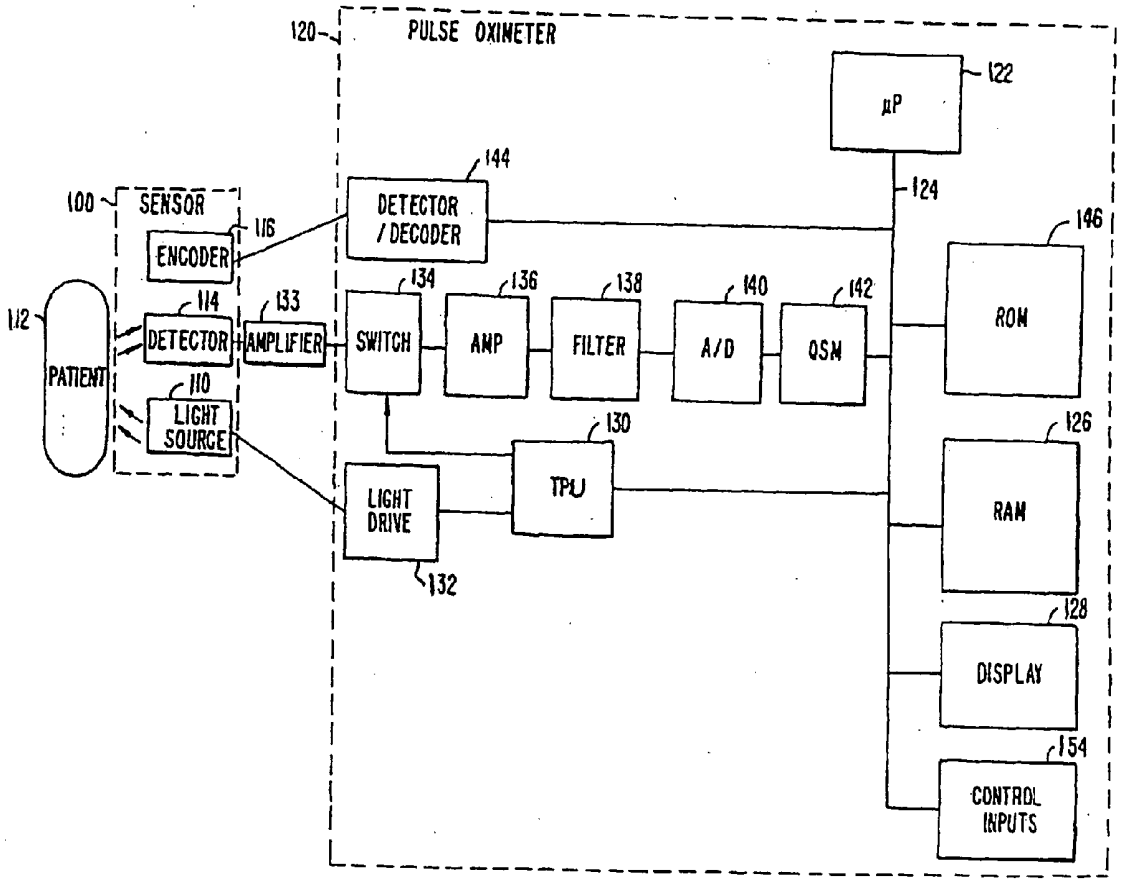
(57) **ABSTRACT**

A method and system for measuring a physiological parameter, comprising collecting a first absorbance at a first wavelength, chosen to be primarily absorbed by water; collecting a second absorbance at a second wavelength, chosen to be primarily absorbed by hemoglobin; and combining the first signal and the second signal to generate a combined plethysmograph signal which is proportionate lower in noise caused by motion-related interference.



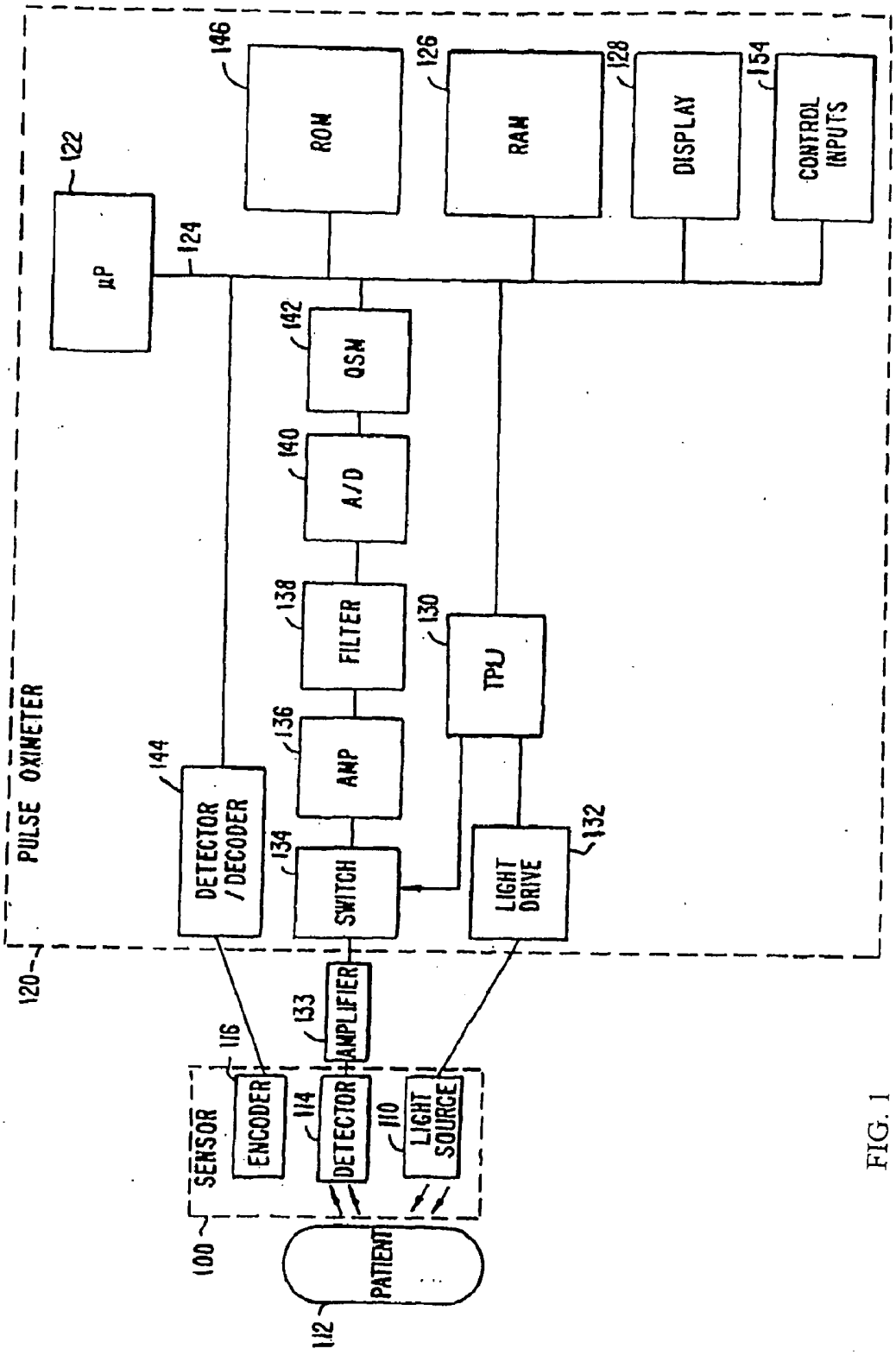

PULSE OXIMETER
120
μP
122
124
ROM
146
RAM
126
DISPLAY
128
CONTROL INPUTS
154
QSM
142
A/D
140
FILTER
138
TPU
130
AMP
136
DETECTOR /DECODER
144
SWITCH
134
LIGHT DRIVE
132
AMPLIFIER
133
116
SENSOR
ENCODER
114
DETECTOR
110
LIGHT SOURCE
100
PATIENT
112

FIG. 1

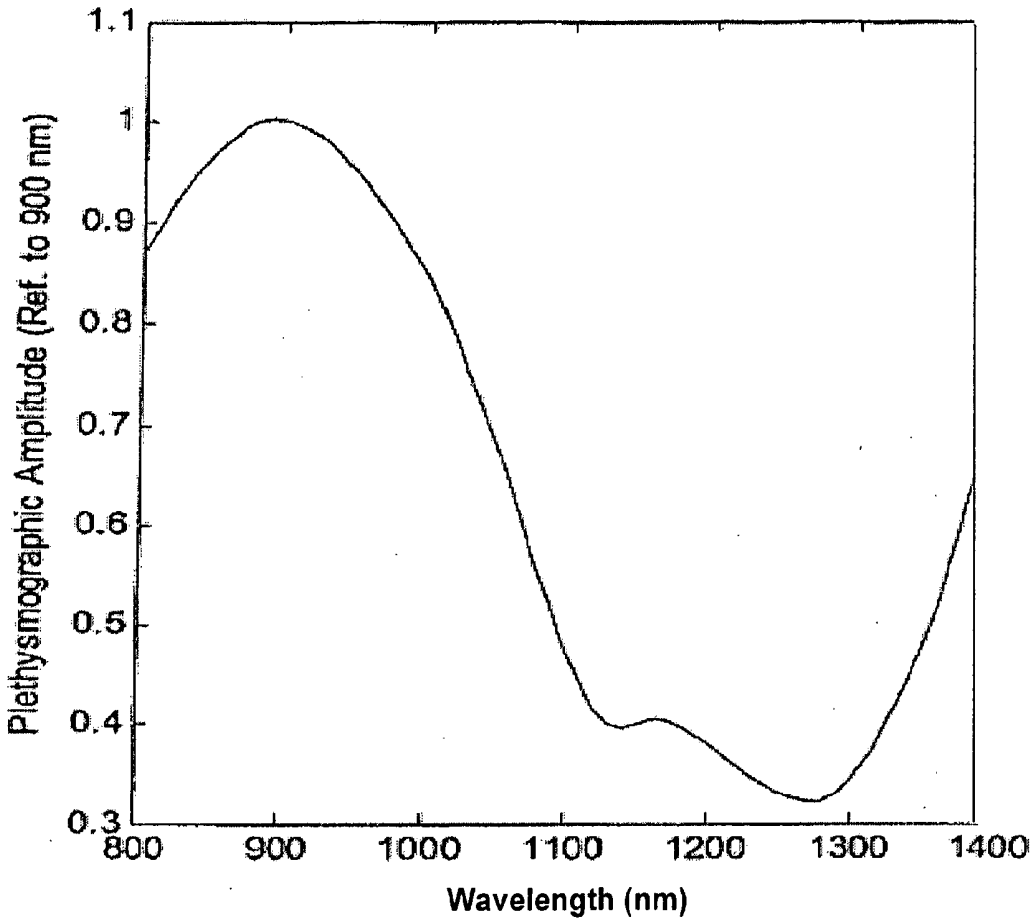

1.1
1
0.9
0.8
0.7
0.6
0.5
0.4
0.3
Plethysmographic Amplitude (Ref. to 900 nm)
800
900
1000
1100
1200
1300
1400
Wavelength (nm)

FIG. 2

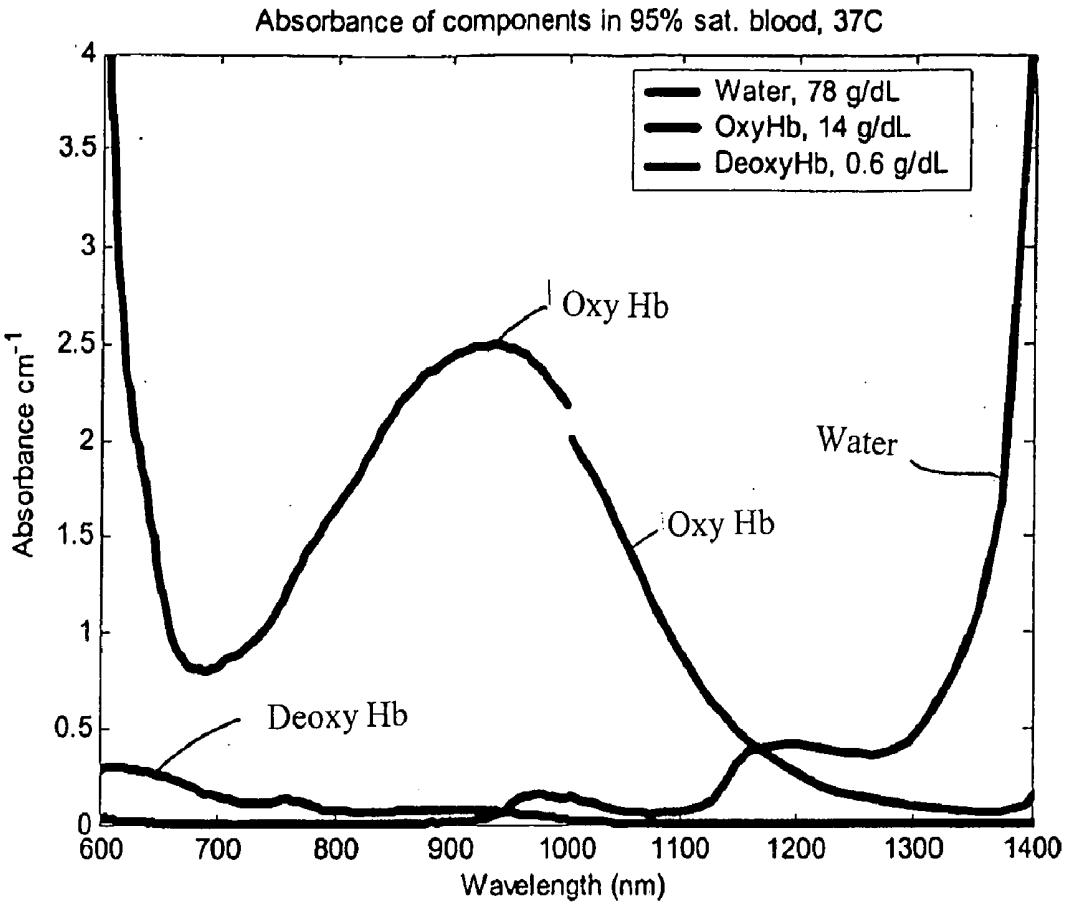

Absorbance of components in 95% sat. blood, 37C
Water, 78 g/dL
OxyHb, 14 g/dL
DeoxyHb, 0.6 g/dL
Oxy Hb
Water
Oxy Hb
Deoxy Hb
Absorbance cm-1
Wavelength (nm)
0
0.5
1
1.5
2
2.5
3
3.5
4
600
700
800
900
1000
1100
1200
1300
1400

FIG. 3

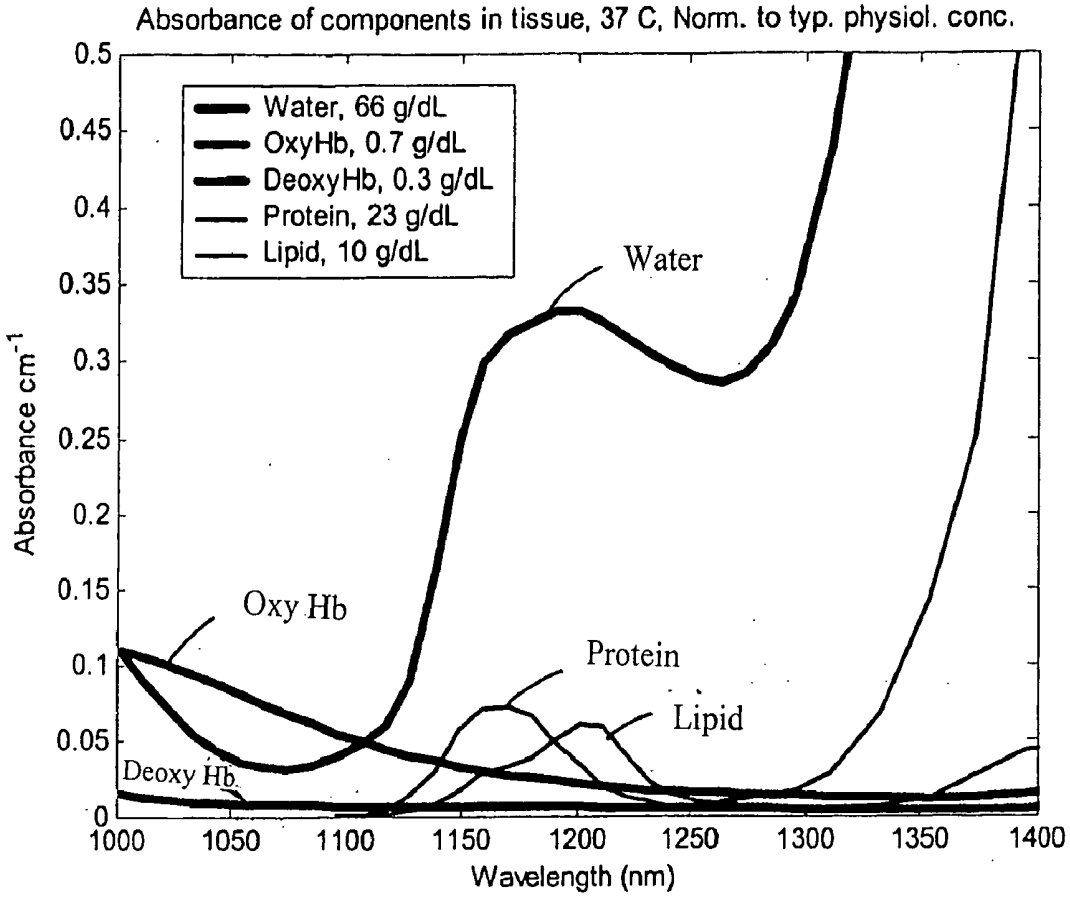

Absorbance of components in tissue, 37 C, Norm. to typ. physiol. conc.
Water, 66 g/dL
OxyHb, 0.7 g/dL
DeoxyHb, 0.3 g/dL
Protein, 23 g/dL
Lipid, 10 g/dL
Water
Oxy Hb
Protein
Lipid
Deoxy Hb
Absorbance cm-1
Wavelength (nm)
0.5
0.45
0.4
0.35
0.3
0.25
0.2
0.15
0.1
0.05
0
1000
1050
1100
1150
1200
1250
1300
1350
1400

FIG. 4

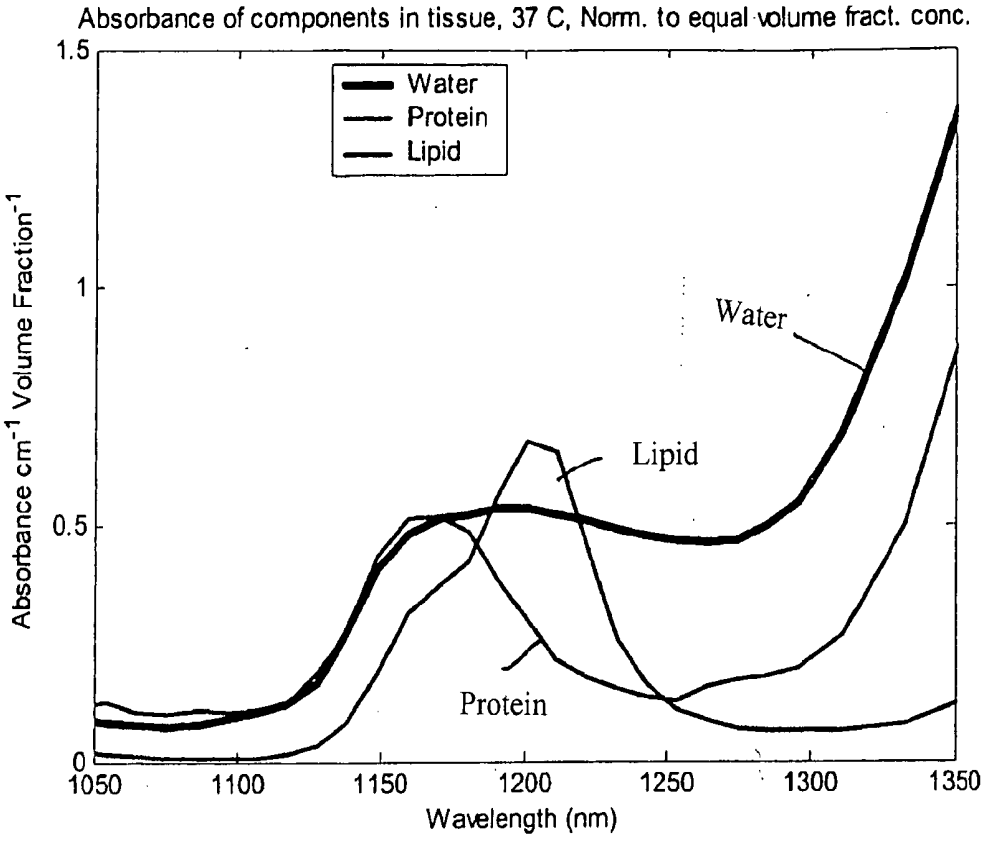

Absorbance of components in tissue, 37 C, Norm. to equal volume fract. conc.
Water
Protein
Lipid
Absorbance cm-1 Volume Fraction-1
1.5
1
0.5
0
1050
1100
1150
1200
1250
1300
1350
Wavelength (nm)
Water
Lipid
Protein

FIG. 5

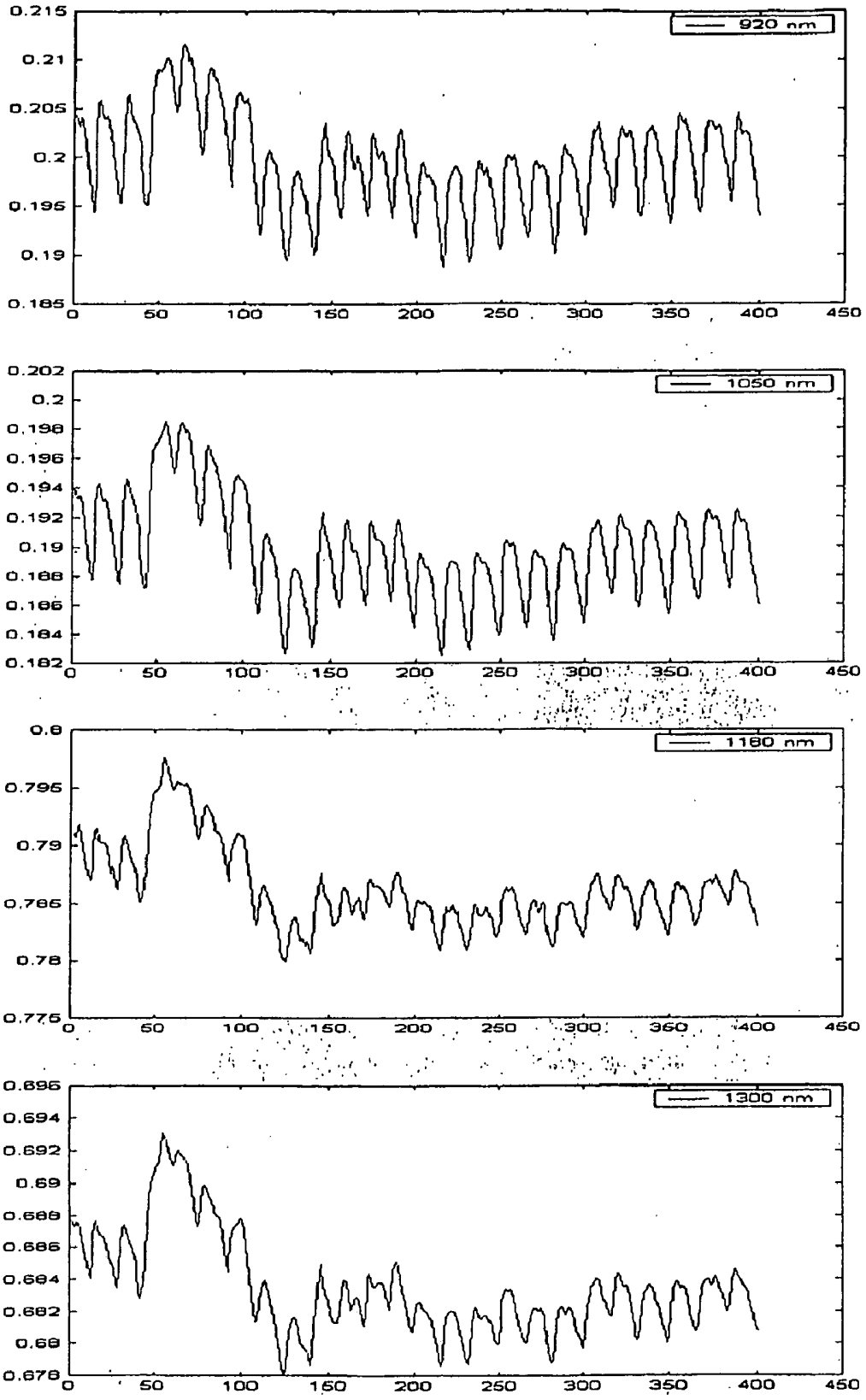


FIG. 6

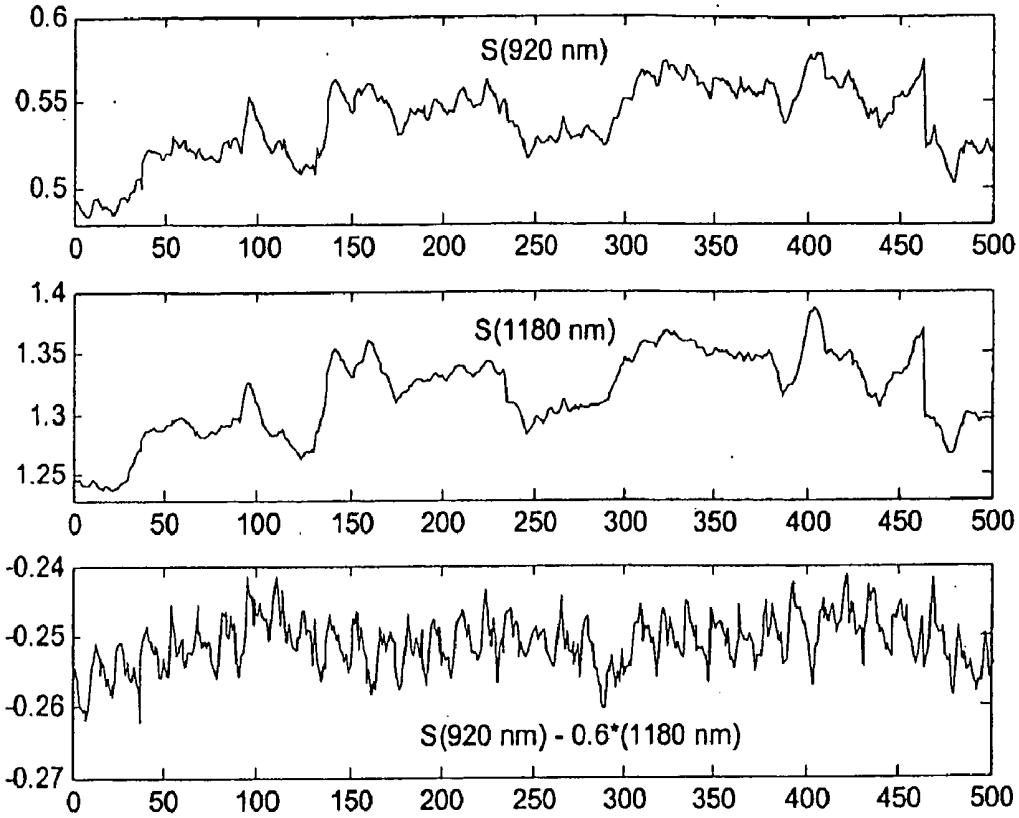

S(920 nm)
S(1180 nm)
S(920 nm) - 0.6*(1180 nm)
0.6
0.55
0.5
1.4
1.35
1.3
1.25
-0.24
-0.25
-0.26
-0.27
0
50
100
150
200
250
300
350
400
450
500

FIG. 7

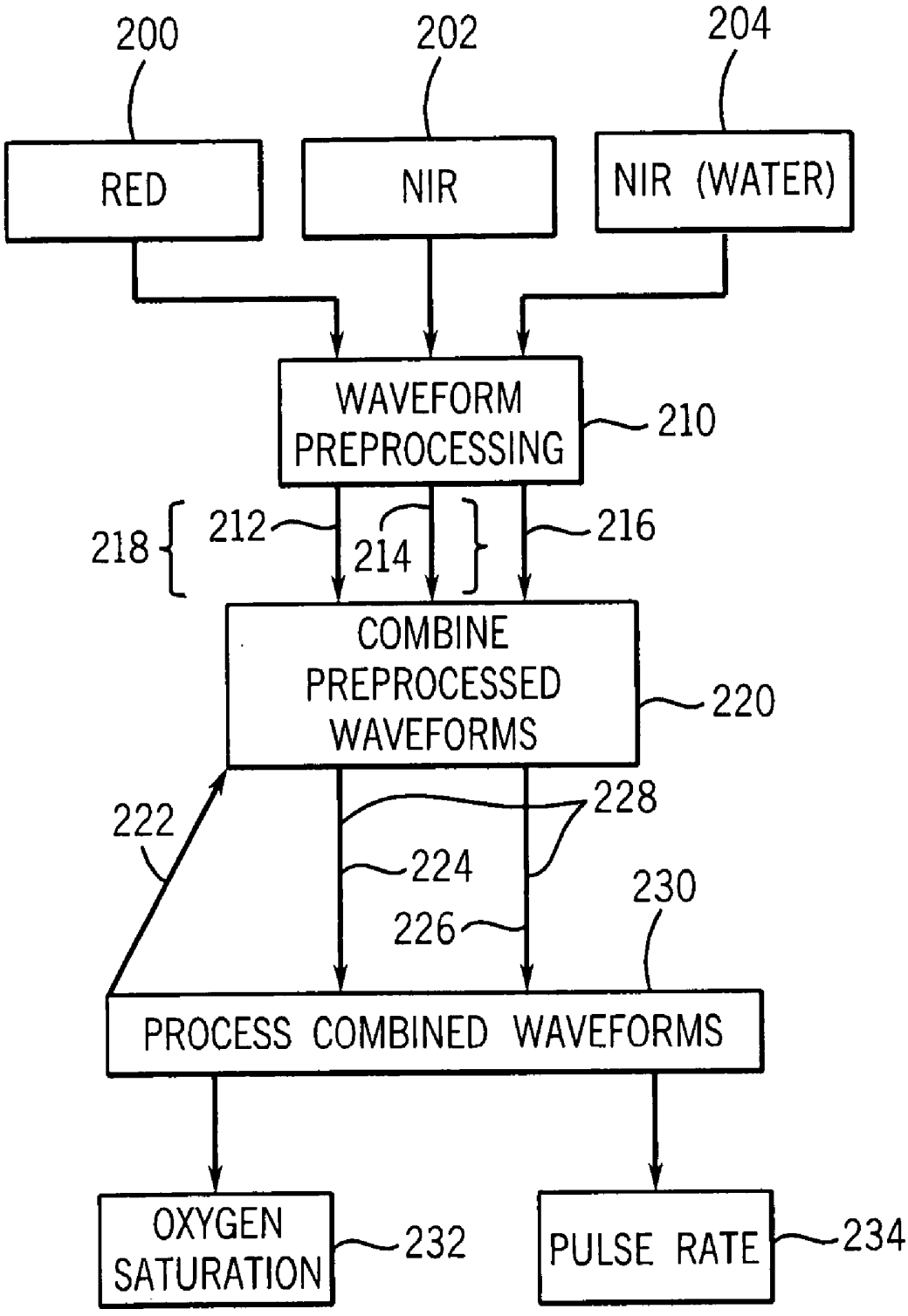

200
202
204
RED
NIR
NIR (WATER)
WAVEFORM PREPROCESSING
210
218
212
214
216
COMBINE PREPROCESSED WAVEFORMS
220
222
228
224
226
230
PROCESS COMBINED WAVEFORMS
OXYGEN SATURATION
232
PULSE RATE
234

FIG. 8

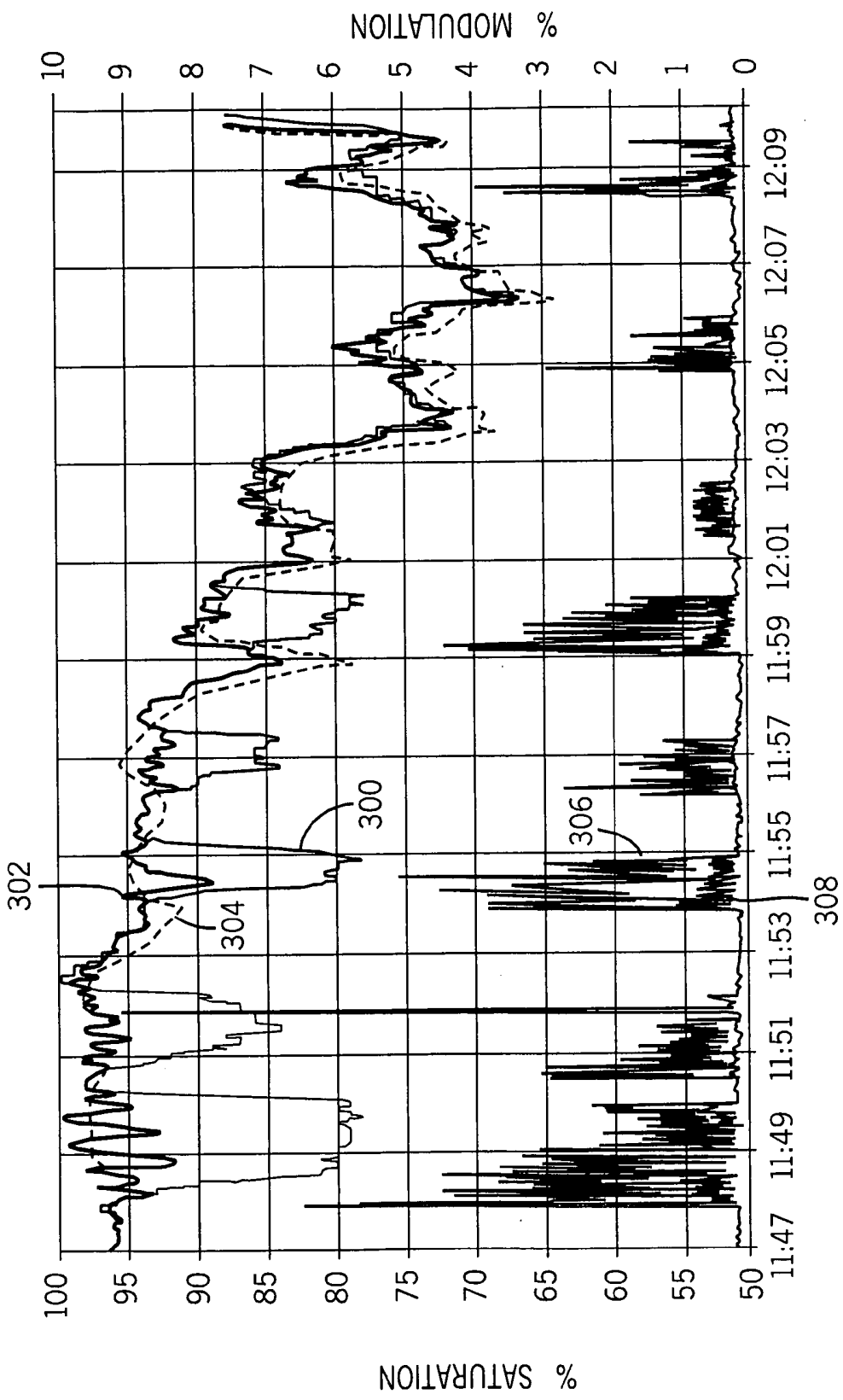

% MODULATION
10
9
8
7
6
5
4
3
2
1
0
11:47
11:49
11:51
11:53
11:55
11:57
11:59
12:01
12:03
12:05
12:07
12:09
% SATURATION
100
95
90
85
80
75
70
65
60
55
50
300
302
304
306
308

FIG. 9

PULSE OXIMETRY SIGNAL CORRECTION USING NEAR INFRARED ABSORPTION BY WATER

CROSS REFERENCE TO RELATED APPLICATIONS

[0001] This application is a continuation-in-part of the U.S. patent Ser. No. 10/797,475, entitled "PULSE OXIMETRY MOTION ARTIFACT REJECTION USING NEAR INFRARED ABSORPTION BY WATER", filed Mar. 9, 2004, which is herein incorporated by reference in its entirety.

BACKGROUND OF THE INVENTION

[0002] 1. Field of the Invention

[0003] The present invention relates generally to the processing of signals obtained from a medical diagnostic apparatus, such a pulse oximeter, using near infrared spectroscopy, to remove artifact or noise effects from the signal representative of a physiological parameter of interest.

[0004] 2. Description of the Related Art

[0005] This section is intended to introduce the reader to various aspects of art that may be related to various aspects of the present invention, which are described and/or claimed below. This discussion is believed to be helpful in providing the reader with background information to facilitate a better understanding of the various aspects of the present invention. Accordingly, it should be understood that these statements are to be read in this light, and not as admissions of prior art.

[0006] A typical pulse oximeter measures two physiological parameters, percent oxygen saturation of arterial blood hemoglobin (SpO_2 or sat) and pulse rate. Oxygen saturation can be estimated using various techniques. In one common technique, the photocurrent generated by the photo-detector is conditioned and processed to determine the ratio of modulation ratios (ratio of ratios) of the red to infrared signals. This modulation ratio has been observed to correlate well to arterial oxygen saturation. The pulse oximeters and sensors are empirically calibrated by measuring the modulation ratio over a range of in vivo measured arterial oxygen saturations (SaO_2) on a set of patients, healthy volunteers, or animals. The observed correlation is used in an inverse manner to estimate blood oxygen saturation (SpO_2) based on the measured value of modulation ratios of a patient. Most pulse oximeters extract the plethysmographic signal having first determined saturation or pulse rate, both of which are susceptible to interference.

[0007] In general, pulse oximetry takes advantage of the fact that in live human tissue, hemoglobin is a strong absorber of light between the wavelengths of 500 and 1100 nm. The pulsation of arterial blood through tissue is readily measurable, using light absorption by hemoglobin in this wavelength range. A graph of the arterial pulsation waveform as a function of time is referred to as an optical plethysmograph. The amplitude of the plethysmographic waveform varies as a function of the wavelength of the light used to measure it, as determined by the absorption properties of the blood pulsing through the arteries. By combining plethysmographic measurements at two different wavelength regions, where oxy- and deoxy-hemoglobin have different absorption coefficients, the oxygen saturation of arterial blood can be estimated. Typical wavelengths employed in commercial pulse oximeters are 660 and 890 nm.

[0008] It is known that rapid motion or application of pressure to a tissue site can have the effect of changing the optical properties being measured at or near that site. The amplitude of the optical signal changes associated with such events, known as motion artifacts, can easily be larger than that due to the arterial pulse. In practice, this can lead to inaccurate estimation of the percent oxygen saturation by pulse oximetry. Various techniques for addressing and removing undesired signal effects, including motion artifacts are known. As used herein, noise refers to signal portions that are undesired or are not directly related to changes in optical properties that are related to the arterial pulse, and which may include motion artifact. The optical signal through the tissue can be degraded by both noise and motion artifact. One source of noise is ambient light which reaches the light detector. Another source of noise is electromagnetic coupling from other electronic instruments. Motion of the patient also introduces noise and affects the signal. For example, the contact between the detector and the skin, or the emitter and the skin, can be temporarily disrupted when motion causes either to move away from the skin. In addition, since blood is a fluid, it responds differently than the surrounding tissue to inertial effects, thus resulting in momentary changes in volume at the point near which the oximeter probe is attached.

[0009] Motion artifact can degrade a pulse oximetry signal relied upon by a health care provider, without the provider's awareness. This is especially true if the monitoring of the patient is remote, the motion is too small to be observed, or the health care provider is watching the instrument or other parts of the patient, and not the sensor site. There are various known techniques for addressing the effects of noise and/or motion artifacts.

[0010] For example, U.S. Pat. No. 4,714,341 discloses a method for combining three wavelengths to detect the presence of motion. The wavelengths are used two at a time to separately compute the oxygen saturation percentage. When the oxygen saturation values computed using different wavelength combinations are in poor agreement, this is assumed to be caused by motion artifact, and the value is discarded. A disadvantage of this approach is that the agreement or lack thereof between the saturation values may or may not be due to motion artifact. In addition, this approach does not identify or remove the effects of motion artifact, but instead discards values that appear suspect

[0011] Another approach involves the filtering of pulse oximetry signals. However, filtering methods require assumptions about the properties of the artifact that do not always hold in practice. In addition, this approach does not measure the motion-induced signal.

[0012] U.S. Pat. No. 5,482,036 provides another approach, and describes a signal processing method for artifact reduction that functions when the artifact-related signal is associated with blood that is at a lower oxygen saturation than the arterial blood. Such a method relies on the generation of an artificial noise signal, which is combined with the physiological parameter to reduce the effect of the unknown noise signal. This approach for reducing the effects of artifact, without separately measuring the motion signal, is based on

assumptions about the effect of motion on the plethysmographic signal. Assumptions may or may not be true, and many assumptions are invalid

[0013] Each of the known techniques for compensating for motion artifact has its own limitations and drawbacks. It is therefore desirable that a pulse oximetry system be designed which more effectively and accurately reports blood-oxygen levels during periods of motion. While many have attempted to isolate the effects of undesired signal portions, such as motion-induced artifacts, by making potentially invalid assumptions or by rejecting suspect estimates of desired signal values, there still remains a need for a deterministic identification, determination and measurement of artifact signals, to enable an accurate measurement of the desired signal values in the presence of undesired signal portions.

SUMMARY

[0014] Certain aspects commensurate in scope with the originally claimed invention are set forth below. It should be understood that these aspects are presented merely to provide the reader with a brief summary of certain forms of the invention might take and that these aspects are not intended to limit the scope of the invention. Indeed, the invention may encompass a variety of aspects that may not be set forth below.

[0015] By measuring the artifact signal, the present technique allows motion artifacts to be separated from the plethysmographic signal without the limiting assumptions of prior known techniques. The present technique provides methods for measuring the motion signal associated with changes in tissue optical properties and using the measurement to compensate plethysmographic measurements made at other wavelengths.

[0016] In one embodiment, the present technique provides a method of determining a physiological parameter, including measuring an absorbance at a wavelength chosen to be primarily absorbed by water, and measuring an absorbance at a wavelength chosen to be primarily absorbed by hemoglobin. A ratio-of-ratios is calculated between these absorbances, and the ratio-of-ratios is used to identify motion noise.

[0017] In another embodiment, there is provided a system for the minimization of motion noise artifacts in pulse oximetry. This system uses a pulse oximeter monitor configured to analyze an absorbance signal that is primarily reflective of motion noise, an absorbance signal chosen to be primarily absorbed by hemoglobin, and another absorbance signal chosen to be primarily absorbed by hemoglobin. In another aspect, the monitor may be configured to use the first and second absorbances to obtain a metric that identifies the presence of motion noise.

[0018] In another embodiment, there is provided one or more tangible, machine readable media, containing code which controls the measurement of an absorbance at a wavelength chosen to be primarily absorbed by water and the measurement of an absorbance at a wavelength chosen to be primarily absorbed by hemoglobin. Under the control of this code, a ratio-of-ratios is calculated between these absorbances, and the ratio-of-ratios is used to identify motion noise.

BRIEF DESCRIPTION OF THE DRAWINGS

[0019] Advantages of the invention may become apparent upon reading the following detailed description and upon reference to the drawings in which:

[0020] FIG. **1** is a block diagram of an exemplary oximeter, in accordance with aspects of the present technique.

[0021] FIG. **2** is a graph of the plethysmographic amplitude measured on the human ear as a function of wavelength.

[0022] FIG. **3** is a graph of absorption spectra of the principal components in human blood, scaled to typical physiological concentration.

[0023] FIG. **4** is a graph of absorption spectra of the principal components in human skin, scaled to typical physiological concentration.

[0024] FIG. **5** is a graph of absorption spectra of the principal components in human skin, scaled to equal volume-fraction concentration.

[0025] FIG. **6** is a graph of plethysmographs measured on a human ear at 4 different wavelengths of approximately 920, 1050, 1180 and 1300 nm respectively.

[0026] FIG. **7** is a graph of an exemplary plethysmographic artifact reduction by combining measurements at 2 near infrared wavelengths.

[0027] FIG. **8** is a flowchart of one approach to using a third wavelength to compensate for motion artifacts, in accordance with aspects of the present technique.

[0028] FIG. **9** is a graph of the oximetry results showing the error compensation using the recombination technique.

DETAILED DESCRIPTION OF SPECIFIC EMBODIMENTS

[0029] One or more specific embodiments of the present invention will be described below. In an effort to provide a concise description of these embodiments, not all features of an actual implementation are described in the specification. It should be appreciated that in the development of any such actual implementation, as in any engineering or design project, numerous implementation-specific decisions must be made to achieve the developers' specific goals, such as compliance with system-related and business-related constraints, which may vary from one implementation to another. Moreover, it should be appreciated that such a development effort might be complex and time consuming, but would nevertheless be a routine undertaking of design, fabrication, and manufacture for those of ordinary skill having the benefit of this disclosure.

[0030] By measuring the artifact signal, the present technique allows motion artifact to be separated from the plethysmographic signal without the limiting assumptions of prior known techniques. The present technique provides methods for measuring the motion signal associated with changes in tissue optical properties and using the measurement to compensate plethysmographic measurements made at other wavelengths.

[0031] FIG. **1** is a block diagram of an exemplary pulse oximeter that may be configured to implement the embodiments of the present technique. The embodiments of the

present technique can be a data processing algorithm that is executed by the microprocessor **122**, described below. Light from light source **110** passes into patient tissue **112**, and is scattered and detected by photodetector **114**. A sensor **100** containing the light source and photodetector may also contain an encoder **116** which provides signals indicative of the wavelength of light source **110** to allow the oximeter to select appropriate calibration coefficients for calculating oxygen saturation. Encoder **116** may, for instance, be a resistor.

[**0032**] Sensor **100** is connected to a pulse oximeter **120**. The oximeter includes a microprocessor **122** connected to an internal bus **124**. Also connected to the bus are a RAM memory **126** and a display **128**. A time processing unit (TPU) **130** provides timing control signals to light drive circuitry **132** which controls when light source **110** is illuminated, and if multiple light sources are used, the multiplexed timing for the different light sources. TPU **130** also controls the gating-in of signals from photodetector **114** through an amplifier **133** and a switching circuit **134**. These signals are sampled at the proper time, depending upon which of multiple light sources is illuminated, if multiple light sources are used. The received signal is passed through an amplifier **136**, a low pass filter **138**, and an analog-to-digital converter **140**. The digital data is then stored in a queued serial module (QSM) **142**, for later downloading to RAM **126** as QSM **142** fills up. In one embodiment, there may be multiple parallel paths of separate amplifiers, filters and A/D converters for multiple light wavelengths or spectra received.

[**0033**] Based on the value of the received signals corresponding to the light received by photodetector **114**, microprocessor **122** will calculate the oxygen saturation using various algorithms. These algorithms require coefficients, which may be empirically determined, corresponding to, for example, the wavelengths of light used. These are stored in a ROM **146**. In one embodiment of a two-wavelength system, the particular set of coefficients chosen for any pair of wavelength spectra is determined by the value indicated by encoder **116** corresponding to a particular light source in a particular sensor **100**. In one embodiment, multiple resistor values may be assigned to select different sets of coefficients. In another embodiment, the same resistors are used to select from among the coefficients appropriate for an infrared source paired with either a near red source or far red source. The selection between whether the near red or far red set will be chosen can be selected with a control input from control inputs **154**. Control inputs **154** may be, for instance, a switch on the pulse oximeter, a keyboard, or a port providing instructions from a remote host computer. Furthermore, any number of methods or algorithms may be used to determine a patient's pulse rate, oxygen saturation or any other desired physiological parameter. For example, the estimation of oxygen saturation using modulation ratios is described in U.S. Pat. No. 5,853,364, entitled "METHOD AND APPARATUS FOR ESTIMATING PHYSIOLOGICAL PARAMETERS USING MODEL-BASED ADAPTIVE FILTERING," issued Dec. 29, 1998, and U.S. Pat. No. 4,911,167, entitled "METHOD AND APPARATUS FOR DETECTING OPTICAL PULSES," issued Mar. 27, 1990, both of which are incorporated herein by reference in their entirety. Furthermore, the relationship between oxygen saturation and modulation ratio is further described in U.S. Pat. No. 5,645,059, entitled "MEDICAL SENSOR WITH MODULATED ENCODING SCHEME," issued Jul. 8, 1997 and incorporated herein by reference in its entirety.

[**0034**] Having described an exemplary pulse oximeter above, the methods for reducing noise, including motion artifact effects in the received signals, according to embodiments of the present technique, are described below.

[**0035**] FIG. **2** is a plot of the average plethysmographic amplitude as a function of wavelength measured through the earlobe of 36 subjects, and normalized to measurements at a wavelength of approximately 900 nm. Measurements, such as those shown in FIG. **2**, reveal that the amplitude of the photoplethysmographic waveform diminishes as a function of wavelength between approximately 900 and 1300 nm, having a minimum value at approximately 1285 nm. The inventors herein have discovered that at wavelengths beyond approximately 900-920 nm, water, which is at much higher concentrations than hemoglobin, also becomes a major light absorber in tissue. FIG. **3** is a graph of some of the light absorbing components found in blood at typical concentrations, in units of absorbance in cm^{-1} vs. wavelength in nm. FIG. **3** shows that at approximately 1300 nm, blood should have only about 20% as much total absorbance as at 900 nm, with water being the dominant absorber. This theoretical model is in rough agreement with the pooled data shown in FIG. **2**, where average plethysmographic amplitude was about ⅓ as much at 1300 nm as at 900 nm.

[**0036**] FIG. **4** is a graph of absorption spectra (cm^{-1}) of the principal components in human skin, scaled to typical physiological concentration, as a function of wavelength in nm. This figures shows that the absorbance due to water has a peak value at approximately 1180 nm, and that similar peaks are present for protein at slightly above 1150 and for lipids at approximately 1200 mm.

[**0037**] FIG. **5** is a graph of absorption spectra of the principal components in human skin, scaled to equal volume-fraction concentration. This figure shows that at approximately 1185 nm, the volume-fraction scaled absorbance for water, lipids and proteins are approximately equal.

[**0038**] While not being limited to any particular theory, the present inventors have, particularly in plethysmographic data from reflectance sensors, noted a weaker effect of water than would be theoretically predicted from absorption spectra. One potential reason for this effect lies in the fact that hemoglobin is largely confined to the blood vessels, whereas water is present at high concentrations both in the blood vessels and in the surrounding tissue. As a result, the pulse-induced expansion of arterial vessels through a tissue bed results in a localized increase in hemoglobin concentration, but only a small net change in water concentration. To the extent that the water concentration in the blood is equal to the water concentration in tissue, the change in light absorption by water is expected to approach zero.

[**0039**] The embodiments of the present technique exploit the finding that in spectral regions where hemoglobin absorbs weakly and water absorbs strongly, the plethysmograph is more sensitive to motion-related events that perturb tissue than arterial pulsation, compared with spectral regions where hemoglobin is a strong absorber and water is a weak absorber.

[**0040**] The weak magnitude of the plethysmograph in regions of strong water absorption is exploited to enable the

separation of arterial-pulse-related signal from a motion artifact signal. By measuring the optical plethysmograph at a wavelength where water is the dominant absorber, the change in tissue optical properties associated with motion or pressure can be measured, with little interference from the underlying arterial pulsation. Plethysmographs at four near-infrared wavelengths measured through a human ear undergoing occasional motion are shown in FIG. **6**, in absorbance units vs. scaled time (i.e., time per point is 43 ms). At approximately 920 nm, where hemoglobin absorption is strong and water absorption is weak, the plethysmograph contains regular arterial pulsations that are interrupted occasionally by motion-related events. As the wavelength is increased to approximately 1300 nm, where water is the predominant absorber, the arterial pulsations diminish and the measured signal becomes largely due to the motion-related events.

[**0041**] By combining the plethysmograph measured in a spectral region where water is the dominant absorber with a plethysmograph measured where blood is a major absorber, the motion-related signal can be selectively removed. FIG. **7** shows the plethysmograph of a human ear measured at approximately 920 nm, and the result of subtracting a portion of the plethysmograph measured at approximately 1180 nm from that measured at 920 nm. In particular, FIG. **7** shows the plethysmograph of a human ear measured at 920 nm, and the result of subtracting approximately 60% of the plethysmograph measured at approximately 1180 nm from that measured at approximately 920 nm. For different wavelength combinations, other multipliers are used based on the ratios of the absorbance of water as compared to that of oxy-hemoglobin or based on empirical determination(s).

[**0042**] By applying the same analysis to a diverse pool of 36 patients measured in a hospital setting, an average signal to noise increase of a factor of 2 of the plethysmograph at 910 nm was observed. By allowing the multiplier for the 1180 nm plethysmograph to vary between subjects, higher signal to noise improvements are achieved.

[**0043**] Theoretical Model

[**0044**] The derivation below and the alternative description that follows demonstrate mechanisms by which the effect of motion-induced changes in optical scattering on a plethysmograph measured at one wavelength can be compensated by plethysmographic measurement at a second wavelength. These are provided as examples of techniques for reducing motion-induced optical changes, but are not the only mechanisms by which the present technique may function, and thus are not meant to limit the embodiments of the present technique.

[**0045**] A starting point for the analysis is the diffusion theory of light transport in tissue (for example, see "Diffusion Theory of Light Transport", Willem M. Star, in Optical-Thermal Response of Laser-Irradiated Tissue, edited by Ashley J. Welch and Martin J. C. van Gemert, Plenum Press, New York, 1995, pgs. 131-206). In the case where the transport-corrected scattering coefficient, μ'_s, is much larger than the absorption coefficient, μ_a, the diffuse intensity of light, $I(\lambda)$, measured at wavelength, λ, by a detector positioned a distance, l, away from a light source, can be described as follows (for example, see "Measurement of Blood Hematocrit by Dual-Wavelength Near-IR Photoplethysmography", Schmitt, J. M.; Guan-Xiong, G.; Miller, J., SPIE, Vol. 1641, 1992, pgs. 150-161):

$$I(\lambda)\alpha \exp(-l\sqrt{(3\mu_a(\lambda)\mu_s(\lambda))}) \quad \text{(eqn. 1)}$$

[**0046**] For small changes in the absorption coefficient, such as those caused by arterial pulsation, the resulting change in intensity can be described by the derivative of intensity with respect to the absorption coefficient:

$$\frac{\frac{dI(\lambda)}{d\mu_a(\lambda)}}{I(\lambda)} = \frac{AC(\lambda)}{DC(\lambda)} = -l\sqrt{\frac{3\mu'_s(\lambda)}{4\mu_a(\lambda)}}\,\Delta V^{art}\mu_a^{art}(\lambda) \quad \text{(eqn. 2)}$$

[**0047**] where ΔV^{art} is the fractional volume change due to arterial pulsation, μ_a^{art} is the absorption coefficient of the arterial blood under measurement, $AC(\lambda)$ refers to the time varying portion of the optical signal and $DC(\lambda)$ refers to the average or non-time varying portion of the optical signal.

[**0048**] The arterial oxygen saturation, SpO_2, is estimated if the AC-DC ratio described by equation 2 is measured at two wavelengths, λ_1 and λ_2, that are chosen so that oxy- and deoxy-hemoglobin are readily differentiated (e.g., λ_1~approximately 660 nm, λ_2.~.approximately 910 nm):

$$R = \frac{\frac{AC(\lambda_1)}{DC(\lambda_1)}}{\frac{AC(\lambda_2)}{DC(\lambda_2)}} = \Omega_{12}\frac{\mu_a^{art}(\lambda_1)}{\mu_a^{art}(\lambda_2)} \quad \text{(eqn. 3a)}$$

where:

$$\Omega_{12} = \sqrt{\frac{\mu'_s(\lambda_1)\mu_a(\lambda_2)}{\mu'_s(\lambda_2)\mu_a(\lambda_1)}} \quad \text{(eqn. 3b)}$$

from which:

$$SpO_2 = \frac{\mu_a^{HHb}(\lambda_1) - R\Omega_{12}^{-1}\mu_a^{HHb}(\lambda_2)}{R\Omega_{12}^{-1}\left(\mu_a^{O_2Hb}(\lambda_2) - \mu_a^{HHb}(\lambda_2)\right) + \mu_a^{HHb}(\lambda_1) - \mu_a^{O_2Hb}(\lambda_1)} \quad \text{(eqn. 3c)}$$

[**0049**] where μ_a^{HHb} and $\mu_a^{O_2Hb}$ are the respective absorption coefficients for deoxy- and oxy-hemoglobin in arterial blood, and R is the ratio of the AC to DC ratios.

[**0050**] The effect of small changes in the scattering coefficient, such as may be brought about by compression of tissue or motion artifact, are as set forth below by eqn. 4:

$$\frac{\frac{dI(\lambda)}{d\mu'_s}}{I(\lambda)} = \frac{AC(\lambda)}{DC(\lambda)} = -l\sqrt{\frac{3\mu_a(\lambda)}{4\mu'_s(\lambda)}}\,\Delta\mu'_s(\lambda) \quad \text{(eqn. 4)}$$

[**0051**] By measuring the AC-DC ratio at a third wavelength, λ_3, chosen so that the absorption due to hemoglobin

is weak but the absorption due to water is strong, the effect of the motion-induced scattering change are removed from the AC-DC measurement at λ_2 by subtracting the scaled AC-DC measurement at λ_3. The resulting motion-corrected plethysmograph, P, can be expressed as:

$$P = \frac{AC(\lambda_2)}{DC(\lambda_2)} - \frac{AC(\lambda_3)}{DC(\lambda_3)}\Omega_{23}^{-1} \quad \text{(eqn. 5a)}$$

where:

$$\Omega_{23} = \sqrt{\frac{\mu_s'(\lambda_2)\mu a(\lambda_3)}{\mu_s'(\lambda_3)\mu_a(\lambda_2)}} \quad \text{(eqn. 5b)}$$

[0052] When the effects of arterial pulsation (equation 2) and motion artifact (equation 4) are additive, equation 5 is expanded as follows:

$$P = -l\sqrt{\frac{3\mu_s'(\lambda_2)}{4\mu_a(\lambda_2)}}\,\Delta V^{art}\mu_a^{art}(\lambda_2) - l\sqrt{\frac{3\mu_a(\lambda_2)}{4\mu_s'(\lambda_2)}}\,\Delta\mu_s'(\lambda_2) + \Omega_{23}^{-1}\left[l\sqrt{\frac{3\mu_s'(\lambda_3)}{4\mu_a(\lambda_3)}}\,\Delta\mu_a(\lambda_3) + l\sqrt{\frac{3\mu_a(\lambda_3)}{4\mu_s'(\lambda_3)}}\,\Delta\mu_s'(\lambda_3)\right] \quad \text{(eqn. 6)}$$

[0053] When water absorption dominates the absorption of light by tissue at λ_3, and the water concentration in the arteries and surrounding tissue is nearly equal, $\Delta\mu_a(\lambda_3)$ is approximately zero, and equation 6 simplifies to:

$$P = -l\sqrt{\frac{3\mu_s'(\lambda_2)}{4\mu_a(\lambda_2)}}\,\Delta V^{art}\mu_a^{art}(\lambda_2) \quad \text{(eqn. 7)}$$

[0054] Equation 7 depends only on the effect of arterial pulsation at λ_2; the effect of the motion artifact has been removed. In a similar manner the plethysmograph measured at λ_3 may be used to remove the motion effects from the plethysmograph measured at λ_1. The corrected plethysmographs measured at λ_1 and λ_2 may then be combined and used to estimate oxygen saturation, as described, for example, by equation 3.

[0055] Several wavelengths in the range between approximately 900 and 1300 nm and more specifically in the range between approximately 1150 and 1350 nm have been tested and found effective at reducing motion-artifact from plethysmographs measured at approximately 910 nm. Wavelengths at the longer wavelength side of this range have the advantage of weaker absorbance of hemoglobin compared to that of water (for example, see FIGS. **3** and **4**). However, wavelengths at the shorter end of this range have the advantage of reduced variation with changing tissue composition. As can be seen in FIG. **5**, where the major components of tissue have been normalized to equal volume fraction, water, lipid, and non-hemoglobin protein all have approximately equal absorbance at approximately 1185 nm. Therefore the absorbance of tissue at approximately 1185 nm will vary little with changes in the relative concentration of these principal components.

[0056] It is known that the detection of light beyond approximately 1100 nm cannot readily be accomplished with the silicon (Si) detectors that are commonly employed in commercial oximeters. For example, the detector used to collect the data displayed in FIGS. **2-7** employed Indium Gallium Arsenide (InGaAs) as the photosensitive material. The most common type of InGaAs detectors are sensitive to light between approximately 800 and 1700 nm. Therefore, in a pulse oximeter designed in accordance with the embodiments of the present technique, with the conventional wavelengths of 660 and 890 nm, in addition to a new light source that emits at wavelengths that are absorbed strongly by water (such as approximately 1180 nm or approximately between 900-1400 nm), an additional detector(s) is used. One such scheme employs two detectors, one Si and one InGaAs, placed side-by-side. An alternative arrangement uses a collinear ("sandwich") detector containing separate Si and InGaAs layers, such as those commercially available, for example, from the Hamamatsu corporation. Yet another alternate arrangement uses two Si detectors placed symmetrically on either side of an InGaAs detector. Alternately, a germanium detector (Ge) is used as a substitute for the InGaAs detector.

[0057] An Implementation of Motion Noise Reduction Technique

[0058] A practical technique by which the effect of motion-induced changes in optical scattering on a plethysmograph measured at one wavelength may be reduced by plethysmographic measurement at a second wavelength is described below. This technique and the derivation above should be considered examples, and are not the only mechanisms by which the present technique may function. They are not meant to limit the embodiments of the present technique.

[0059] In one example, three wavelengths of light are used: a red wavelength at 660 nm, a near infrared (NIR) wavelength at 890 nm, and a NIR wavelength at 1300 nm. The first two wavelengths are both chosen to be primarily absorbed by hemoglobin, and the third wavelength is chosen to be primarily absorbed by water. After these wavelengths of light from the light source **110** (See FIG. **1**) are passed through the tissue, the light is collected by a photodetector **114** (See FIG. **1**) generating plethysmographs at each frequency.

[0060] Turning now to FIG. **8**, in an exemplary embodiment, the red plethysmograph **200**, the NIR plethysmograph **202**, and the NIR (water) plethysmograph **204** are preprocessed (Block **210**) prior to use. In this step, the waveforms are converted to a natural logarithm, and may by filtered to reduce noise, such as with a bandpass filter. The preprocessed plethysmographs **218** are then mathematically combined (Block **220**) to identify periods of high motion noise and to generate plethysmographs with reduced motion noise.

[0061] In one such embodiment, the preprocessed NIR plethysmograph **214** and the preprocessed NIR (water) plethysmograph **216** are used to identify periods of high and/or low motion noise. This is performed by calculating a

ratio-of-ratios, $R_{1300,890}$, between the absorbances at the NIR wavelength (890 nm) and the water wavelength (1300 nm) (See Eqn. 3a above for an example). The value of this ratio is less than 1.0 for periods when there are little or no motion artifacts, ranging from around 0.2 to 0.7 for most subjects. In one embodiment, a default value of 0.4 may be selected for initial use by an algorithm as described herein. In an exemplary embodiment, $R_{1300,890}$ is calculated on two periods: once using three seconds of data for rapid detection of motion artifacts, and once using fifteen seconds of data for use in adjusting the combined weights, as discussed further below.

[0062] In one embodiment, a three step process is used to generate plethysmographs with reduced motion noise. The first step is to subtract fractions (F) of the preprocessed NIR (water) plethysmograph **216** ($Preprocessed_{1300}$) from the preprocessed red plethysmograph **212** ($Preprocessed_{660}$), and the preprocessed NIR plethysmograph **214** ($Preprocessed_{890}$), to generate corrected waveforms:

$$Corrected_{890}=Preprocessed_{890}-F_{1300,890}*Preprocessed_{1300} \quad \text{(eqn. 8)}$$

$$Corrected_{660}=Preprocessed_{660}-F_{1300,660}*Preprocessed_{1300} \quad \text{(eqn. 9)}$$

[0063] In such an embodiment, the second step is to rescale the corrected waveforms to preserve the ratio-of-ratios **222** ($R_{660,890}$) between the absorbance signals at the red (660 nm) and NIR (890 nm) wavelengths, so that the coefficients in eqn. 3b will not need to change. This is performed by estimating the fractions (C_{890} and C_{660}) of the arterial pulse that were cancelled in $Corrected_{890}$ and $Corrected_{660}$:

$$Rescaled_{890}=Corrected_{890}/(1.0-C_{890}) \quad \text{(eqn. 10)}$$

$$Rescaled_{660}=Corrected_{660}/(1.0-C_{660}) \quad \text{(eqn. 11)}$$

where:

$$C_{890}=R_{1300,890}*F_{1300,890} \quad \text{(eqn. 12)}$$

$$C_{660}=R_{1300,890}*F_{1300,660} \quad \text{(eqn. 13)}$$

$R_{660,890}$ **222** may be supplied by the two wavelength oximetry algorithm **230**, which calculates this value for determination of the oxygen saturation. An alternative method for rescaling $Corrected_{660}$ is to add a percentage of $Corrected_{890}$ to maintain a constant value for $R_{660,890}$:

$$Rescaled_{660}=Corrected_{660}+(F_{1300,660}/F_{1300,890})*C_{890}*Rescaled_{890} \quad \text{(eqn. 14)}$$

[0064] In this embodiment, the third step in generating plethysmographs with reduced motion noise is to adjust the fractions, $F_{1300,660}$ and $F_{1300,890}$, of the NIR (water) plethysmograph **216** ($Preprocessed_{1300}$) subtracted from the other two waveforms. Mathematical techniques may be selected that minimize the power, standard deviation, or amplitude of the resulting waveforms. Alternatively, techniques may be chosen that minimize the skewness of the derivative of the rescaled waveforms, or enhance some other recognized metric, or combination of metrics, of signal quality. The techniques for adjusting the fractions, $F_{1300,660}$ and $F_{1300,890}$ may be selected based on their efficacy in reducing saturation or pulse rate errors in representative sets of oximetry data that include motion artifact.

[0065] An example of one technique for calculating $F_{1300,660}$ and $F_{1300,890}$, is to use the summations given below:

$$F_{1300,890}=\frac{\sum Preprocessed_{890,t}Preprocessed_{1300,t}}{\sum Preprocessed^2_{1300,t}}-\frac{\sum R_{1300,890}(1.0-C_{890})Rescaled^2_{890,t}}{\sum Preprocessed^2_{1300,t}} \quad \text{(eqn. 15)}$$

$$F_{1300,660}=\frac{\sum Preprocessed_{660,t}Preprocessed_{1300,t}}{\sum Preprocessed^2_{1300,t}}-\frac{\sum R_{1300,890}(1.0-C_{890})R_{660,890}Rescaled^2_{890,t}}{\sum Preprocessed^2_{1300,t}} \quad \text{(eqn. 16)}$$

These summations may be adequately represented by the approximations shown below:

$$F_{1300,890}=\frac{\sum Preprocessed_{890,t}Preprocessed_{1300,t}}{\sum Preprocessed^2_{1300,t}}-\varepsilon \quad \text{(eqn. 17)}$$

$$F_{1300,660}=\frac{\sum Preprocessed_{660,t}Preprocessed_{1300,t}}{\sum Preprocessed^2_{1300,t}}-\varepsilon \quad \text{(eqn. 18)}$$

In one implementation using these summations, a value of 0.03 has been found to work well for ϵ. Alternatively, these summations may be approximated with infinite impulse response (IIR) filters. The values for $F_{1300,660}$ and $F_{1300,890}$ typically range from 0.6-0.9, and, in one embodiment, may be limited to range between 0.5-1.0 with a default value of 0.7. As will be understood by those skilled in the art, these constants may vary due to factors such as wavelength selection or sensor site or geometry.

[0066] As shown in FIG. **8**, the plethysmographs which have been adjusted to reduce the noise motion artifacts, $Rescaled_{890}$ **226** and $Rescaled_{660}$ **224**, are then used in a two wavelength algorithm **230** to calculate a value for oxygen saturation **232** and pulse rate **234**. In one embodiment, the two wavelength algorithm **230** may be similar to that described in U.S. Pat. No. 5,853,364, but without the preprocessing that has already been done in block **210** of FIG. **8**.

[0067] The improvements afforded by this technique are illustrated in the graph shown in FIG. **9**. For this test, an oxygen sensor was attached to a test subject's ear lobe, which is highly susceptible to motion artifacts. As a control, another sensor was attached to a digit on the test subject. This second sensor was connected to a oximeter using a standard two wavelength algorithm. In the graph, the oxygen saturation **300** calculated from the preprocessed red plethysmograph **212** (See FIG. **8**) and the preprocessed NIR plethysmograph **214** (See FIG. **8**), using a standard two wavelength algorithm, showed a significant drop during periods of motion, such as nodding or shaking of the head. A control value **304** was calculated from the sensor located on the digit and remained steady. In contrast to the oxygen saturation **300** calculated from the uncombined preprocessed plethysmographs **218** (See FIG. **8**), the oxygen

saturation **302** calculated from the combined plethysmographs **228** (See FIG. **8**) closely tracked the control. This is further illustrated by the % modulation curves at the bottom of the graph in FIG. **9**. Prior to correction, the % modulation signal **306** shows the motion noise to be far larger then the % modulation signal **308** after the technique above is used.

[**0068**] In a larger test, a test group of 10 subjects using the standard two wavelength algorithm showed a pooled root-mean-square-difference (RMSD) in oxygen saturation of 4.55%, with some periods of 25% errors, between the moving sensor and a non-moving control. In contrast, the same data processed by the three wavelength algorithm discussed above showed a RMSD of 2.61% for the pooled subjects.

[**0069**] The values calculated in the algorithm detailed above may be used in a number of ways to display more accurate information to the user, while minimizing the load on the processor. For example, turning back to FIG. **8**, during periods of very low motion artifacts, such as where $R_{1300,890}<0.85$, the calculation above may be deactivated and the oxygen saturation **232** and pulse rate **234** calculated using the uncombined data from the preprocessed plethysmographs **218**. Conversely, in such an embodiment, during periods of high motion artifacts, the calculation may remain active or may be activated and the oxygen saturation **232** and pulse rate **234** calculated using the combined data from the preprocessed plethysmographs **218**. Alternatively, the value for $R_{1300,890}$ could be used to gradually interpolate between values calculated from the preprocessed plethysmographs **218** and the combined plethysmographs **228**. This technique would be useful in cases where the NIR (water) plethysmograph **204** was weaker, perhaps due to small pulse amplitude or a thick sensor site. In this case, the NIR (water) plethysmograph **204** would have a poor signal-to-noise ratio, and using the combined plethysmographs **228** only during periods of high motion artifacts would provide the most accurate information.

[**0070**] Additional useful modifications could take advantage of the extra signals provided by the technique. For example, additional preprocessing filters may be implemented prior to the calculation of the adjusted waveforms. In another example, various algorithms in the oximeter, such as sensor off detection, may continue to use the preprocessed plethysmographs **218**, while the oxygen saturation and pulse rate calculation use the combined plethysmographs **228**.

[**0071**] In addition, an alternative wavelength selection to the above-described augmentation to conventional pulse oximetry is an all-NIR pulse oximeter. An example of an all NIR oximeter is an oximeter employing light sources emitting at approximately 940, 1040, and 1180 nm used in conjunction with a single InGaAs detector. In addition to the advantage of requiring only one detector, the all-NIR implementation has advantages associated with the optical properties of tissue. The accuracy of measurements made using pulse oximetry depends, in part, on the extent to which the paths traveled by the different colors of light are the same. The mean path length and penetration depth of light at a particular wavelength traveling through tissue is strongly affected by the absorption and scattering coefficients of tissue at that wavelength. In conventional pulse oximetry, in order to achieve the same mean path length and penetration depth at two wavelengths, the scattering and absorption coefficients at the two wavelengths need to be matched. The scattering of light by tissue decreases rapidly as a function of wavelength, with the result that the scattering properties of tissue at approximately 940, 1040, and 1180 nm will be more closely matched than the scattering properties of tissue at a combination of both visible and NIR wavelengths such as approximately 660, 890, and 1180 nm, for reasons discussed below. The absorption properties of oxy- and deoxy-hemoglobin are such that at high oxygen saturation values the net (i.e., combined effects of oxy and deoxy) absorption coefficient due to hemoglobin will be matched reasonably well at 660 nm and 940 nm. However, as oxygen saturation values decrease, the high absorption coefficient of deoxy-hemoglobin at approximately 660 nm will result in an increasingly strong mismatch between the net absorption coefficient of hemoglobin at approximately 660 and approximately 940 nm. The net absorption coefficients of hemoglobin at approximately 940 and approximately 1040 nm, will be more closely matched than at approximately 660 and approximately 940 nm, over the full range of measurable oxygen saturation values.

[**0072**] The choice of the wavelength used to measure the motion-artifact signal depends partially on the need for matching the optical path length to that of the signals to be corrected. Beyond approximately 950 nm, the absorption coefficient of water, protein, and non-hemoglobin protein, in addition to that of hemoglobin needs to be considered in order to achieve close matching of path lengths. Although about 1300 nm is a currently preferred wavelength for measuring the motion-artifact signal, other alternative wavelength values are also effective, for example, wavelengths between approximately 1050 and 1400 nm and between approximately 1500 and 1850 nm.

[**0073**] The embodiments of the present technique may be practiced by placing the optical components directly at the tissue interface, or alternatively, by transporting the light to and from the tissue with fiber optics. The former implementation has the advantage of more efficient delivery and collection of the light, whereas the latter implementation has the advantages of being less costly. The less costly solution is enabled by the fact that when employing fiber optic delivery, the light sources and detectors can reside in the monitor as opposed to the sensor, and considering that such components may be more expensive that the fiber, this will result in a less expensive device.

[**0074**] As will be understood by those skilled in the art, other equivalent or alternative methods for the measurement of motion artifact signal associated with changes in tissue optical properties, and using the measurement to compensate plethysmographic measurements made at other wavelengths, according to the embodiments of the present technique can be envisioned without departing from the essential characteristics thereof. For example, a combination of visible and NIR or an all NIR wavelength combination may be used to make the measurements. Moreover, individuals skilled in the art of near-infrared spectroscopy would recognize that additional terms can be added to the algorithms used herein to incorporate reflectance measurements made at additional wavelengths and thus improve accuracy further. Also, light sources or light emission optics other then LED's including and not limited to incandescent light and narrow-band light sources appropriately tuned to the desired wavelengths and associated light detection optics may be placed

near the tissue location or may be positioned within a remote unit; and which deliver light to and receive light from the tissue location via optical fibers. Additionally, sensor arrangements functioning in a back-scattering or a reflection mode to make optical measurements of reflectances, as well as other embodiments, such as those working in a forward-scattering or a transmission mode may be used to make these measurements.

[0075] While the invention may be susceptible to various modifications and alternative forms, specific embodiments have been shown by way of example in the drawings and have been described in detail herein. However, it should be understood that the invention is not intended to be limited to the particular forms disclosed. Rather, the invention is to cover all modifications, equivalents, and alternatives falling within the spirit and scope of the invention as defined by the following appended claims.

What is claimed is:

1. A method of determining a physiological parameter, comprising:

obtaining a first absorbance at a first wavelength, wherein the first wavelength is chosen to be primarily absorbed by water;

obtaining a second absorbance at a second wavelength, wherein the second wavelength is chosen to be primarily absorbed by hemoglobin;

estimating a first ratio-of-ratios between the first absorbance and the second absorbance; and

using the first ratio-of-ratios to identify motion noise.

2. The method of claim 1, comprising

obtaining a third absorbance at a third wavelength, wherein the third wavelength is chosen to be primarily absorbed by hemoglobin;

calculating a corrected second absorbance by subtracting a first fraction of the first absorbance from the second absorbance; and

calculating a corrected third absorbance by subtracting a second fraction of the first absorbance from the third absorbance.

3. The method of claim 2, comprising:

resealing the corrected second absorbance to obtain a rescaled second absorbance; and

rescaling the corrected third absorbance to obtain a rescaled third absorbance,

wherein the rescaling of the corrected second absorbance and the corrected third absorbance maintains a ratio-of-ratios between the rescaled second absorbance and the rescaled third absorbance that is independent of changes in the subtracted first and section fractions of the first absorbance.

4. The method of claim 3, further comprising:

calculating a first oxygen saturation value using a two wavelength algorithm based on the rescaled second absorbance and the rescaled third absorbance; and

calculating a second oxygen saturation value using a two wavelength algorithm based on the second absorbance and the third absorbance.

5. The method of claim 3, further comprising:

calculating an oxygen saturation value using a two wavelength algorithm based on the rescaled second absorbance and the rescaled third absorbance.

6. The method of claim 4, further comprising:

displaying the first oxygen saturation value during periods of high motion noise;

displaying the second oxygen saturation value during periods of low motion noise;

calculating an intermediate oxygen saturation value using the first and second oxygen saturation values; and

displaying the intermediate oxygen saturation value during periods of intermediate motion noise.

7. The method of claim 3, further comprising:

calculating a first pulse rate value using the rescaled second absorbance; and

calculating a second pulse rate value using the second absorbance.

8. The method of claim 3, further comprising:

calculating a pulse value using the corrected second absorbance or the rescaled second absorbance.

9. The method of claim 7, further comprising:

displaying the first pulse rate value during periods of high motion noise;

displaying the second pulse rate value during periods of low motion noise;

calculating an intermediate pulse rate value using the first and second pulse rate values; and

displaying the intermediate pulse rate value during periods of intermediate motion noise.

10. The method of claim 2, further comprising using an algorithm to calculate the value of the first fraction and the value of the second fraction.

11. The method of claim 10, wherein the algorithm minimizes the standard deviation of a resulting waveform.

12. The method of claim 10, wherein the algorithm minimizes the power of a resulting waveform.

13. The method of claim 10, wherein the algorithm minimizes the skewness of a resulting waveform.

14. The method of claim 10, wherein the algorithm is chosen to reduce errors in the calculation of oxygen saturation and pulse rate.

15. The method of claim 10, wherein the algorithm comprises an infinite impulse response (IIR) filter.

16. A system for signal correction in pulse oximetry, the system comprising:

a pulse oximeter monitor, wherein the pulse oximeter monitor is configured to analyze a first absorbance signal substantially corresponding to motion noise, to analyze a second absorbance signal chosen to be primarily absorbed by hemoglobin, to analyze a third absorbance signal chosen to be primarily absorbed by hemoglobin, to adjust the second absorbance signal to compensate for noise by subtracting a first fraction of the first absorbance signal from the second absorbance signal to obtain a corrected second absorbance signal, and to adjust the third absorbance signal to compensate for noise by subtracting a second fraction of the first

absorbance signal from the third absorbance signal to obtain a corrected third absorbance signal.

17. The system of claim 16, further comprising a pulse oximetry sensor, wherein the sensor comprises optical emitters and detectors configured to emit and detect light at a first wavelength chosen to be primarily absorbed by water, to emit and detect light at a second wavelength chosen to be primarily absorbed by hemoglobin, and to emit and detect light at a third wavelength chosen to be primarily absorbed by hemoglobin.

18. The system of claim 16, wherein the pulse oximeter monitor is configured to combine the first absorbance signal and the second absorbance signal to obtain a metric that identifies the presence of motion noise.

19. The system of claim 16, wherein the pulse oximeter monitor is configured to rescale the corrected second absorbance signal to obtain a rescaled second absorbance signal, and to rescale the corrected third absorbance signal to obtain a rescaled third absorbance signal, wherein the resealing maintains a ratio-of-ratios between the rescaled second absorbance signal and the rescaled third absorbance signal that is independent of changes in the subtracted first and section fractions of the first absorbance.

20. The system of claim 19, wherein the pulse oximeter monitor is configured to calculate a first oxygen saturation value using the rescaled second absorbance signal and the rescaled third absorbance signal, and to calculate a second oxygen saturation value using the first absorbance signal and the second absorbance signal.

21. The system of claim 19, wherein the pulse oximeter monitor is configured to calculate an oxygen saturation value using the rescaled second absorbance signal and the rescaled third absorbance signal.

22. The system of claim 20, wherein the pulse oximeter monitor is configured to display the first oxygen saturation value during periods of high motion noise, to display the second oxygen saturation value during periods of low motion noise, to calculate an intermediate oxygen saturation value using the first oxygen saturation value and the second oxygen saturation value, and to display the intermediate oxygen saturation value during periods of intermediate motion noise.

23. The system of claim 19, wherein the pulse oximeter monitor is configured to calculate a first pulse rate value from the rescaled second absorbance signal and the rescaled third absorbance signal, and to calculate a second pulse rate value from the second absorbance signal and the third absorbance signal.

24. The system of claim 19, wherein the pulse oximeter monitor is configured to calculate a pulse rate value from the rescaled or corrected second and third absorbance signals.

25. The system of claim 23, wherein the pulse oximeter monitor is configured to display the first pulse rate value during periods of high motion noise, to display the second pulse rate value during periods of low motion noise, to calculate an intermediate pulse rate value using the first pulse rate value and the second pulse rate value, and to display the intermediate pulse rate value during periods of intermediate motion noise.

26. One or more tangible, machine readable media, comprising code executable to perform the acts of:

obtaining a first absorbance at a first wavelength, wherein the first wavelength is chosen to be primarily absorbed by water;

obtaining a second absorbance at a second wavelength, wherein the second wavelength is chosen to be primarily absorbed by hemoglobin;

estimating a first ratio-of-ratios between the first absorbance and the second absorbance; and

using the first ratio-of-ratios to identify motion noise.

27. The one or more tangible, machine readable media of claim 26, further comprising code executable to perform the acts of:

obtaining a third absorbance at a third wavelength, wherein the third wavelength is chosen to be primarily absorbed by hemoglobin;

calculating a corrected second absorbance by subtracting a first fraction of the first absorbance from the second absorbance; and

calculating a corrected third absorbance by subtracting a second fraction of the first absorbance from the third absorbance.

28. The one or more tangible, machine readable media of claim 27, further comprising code executable to perform the acts of:

rescaling the corrected second absorbance to obtain a rescaled second absorbance; and

rescaling the corrected third absorbance to obtain a rescaled third absorbance,

wherein the rescaling of the corrected second absorbance and the corrected third absorbance maintains a ratio-of-ratios between the rescaled second absorbance and the rescaled third absorbance that is independent of changes in the subtracted first and section fractions of the first absorbance.

29. The one or more tangible, machine readable media of claim 28, further comprising code executable to perform the acts of:

calculating a first oxygen saturation value using a two wavelength algorithm based on the rescaled second absorbance and the rescaled third absorbance; and

calculating a second oxygen saturation value using a two wavelength algorithm based on the second absorbance and the third absorbance.

30. The one or more tangible, machine readable media of claim 27, further comprising code executable to perform the acts of:

Calculating an oxygen saturation value using a two wavelength algorithm based on the rescaled second absorbance and the rescaled third absorbance.

31. The one or more tangible, machine readable media of claim 29, further comprising code executable to perform the acts of:

displaying the first oxygen saturation value during periods of high motion noise;

displaying the second oxygen saturation value during periods of low motion noise;

calculating an intermediate oxygen saturation value using the first oxygen saturation value and the second oxygen saturation value; and

displaying the intermediate oxygen saturation value during periods of intermediate motion noise.

32. The one or more tangible, machine readable media of claim 28, further comprising code executable to perform the acts of:

calculating a first pulse rate value using the rescaled second absorbance; and

calculating a second pulse rate value using the second absorbance.

33. The one or more tangible, machine readable media of claim 28, further comprising code executable to perform the acts of:

calculating a pulse rate value using the rescaled second absorbance or the corrected second absorbance.

34. The one or more tangible, machine readable media of claim 32, further comprising code executable to perform the acts of:

displaying the first pulse rate value during periods of high motion noise;

displaying the second pulse rate value during periods of low motion noise;

calculating an intermediate pulse rate value using the first pulse rate value and the second pulse rate value; and

displaying the intermediate pulse rate value during periods of intermediate motion noise.

35. The one or more tangible, machine readable media of claim 27, further comprising code executable to perform the act of using an algorithm to calculate the value of the first fraction and the value of the second fraction.

36. The one or more tangible, machine readable media of claim 35, wherein the algorithm minimizes the standard deviation of a resulting waveform.

37. The one or more tangible, machine readable media of claim 35, wherein the algorithm minimizes the power of a resulting waveform.

38. The one or more tangible, machine readable media of claim 35, wherein the algorithm minimizes the skewness of a resulting waveform.

39. The one or more tangible, machine readable media of claim 35, wherein the algorithm is chosen to reduce errors in the calculation of oxygen saturation and pulse rate.

40. The one or more tangible, machine readable media of claim 35, wherein the algorithm is an infinite impulse response (IIR) filter.

* * * * *

专利名称(译)	利用水的近红外吸收进行脉搏血氧饱和度信号校正		
公开(公告)号	US20070106137A1	公开(公告)日	2007-05-10
申请号	US11/521960	申请日	2006-09-15
[标]申请(专利权)人(译)	BAKER JR CLARK - [R 喀斯特爱德华 HOARAU CARINE		
申请(专利权)人(译)	BAKER CLARK - [R JR 喀斯特爱德华 HOARAU CARINE		
当前申请(专利权)人(译)	COVIDIEN LP		
[标]发明人	BAKER CLARK R JR KARST EDWARD HOARAU CARINE		
发明人	BAKER, CLARK R. JR. KARST, EDWARD HOARAU, CARINE		
IPC分类号	A61B5/00 A61B5/024		
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摘要(译)

一种用于测量生理参数的方法和系统，包括收集第一波长的第一吸光度，选择为主要被水吸收;收集第二波长的第二吸光度，选择主要被血红蛋白吸收;并且组合第一信号和第二信号以产生组合的体积描记器信号，其与由运动相关的干扰引起的噪声成比例地降低。