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(54) **OPTICAL MEASUREMENT APPARATUS AND BLOOD SUGAR LEVEL MEASURING APPARATUS USING THE SAME**

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(57) **ABSTRACT**

An apparatus for non-invasively measuring blood sugar levels based on temperature measurements. A blood sugar level non-invasively measured by a temperature measuring method is corrected by blood oxygen saturation and blood flow volume. Optical sensors detect scattered light, reflected light, and light exiting from a body surface after penetrating the skin, so that measurement data can be stabilized by taking into consideration the influence of the thickness of the skin on blood oxygen saturation.

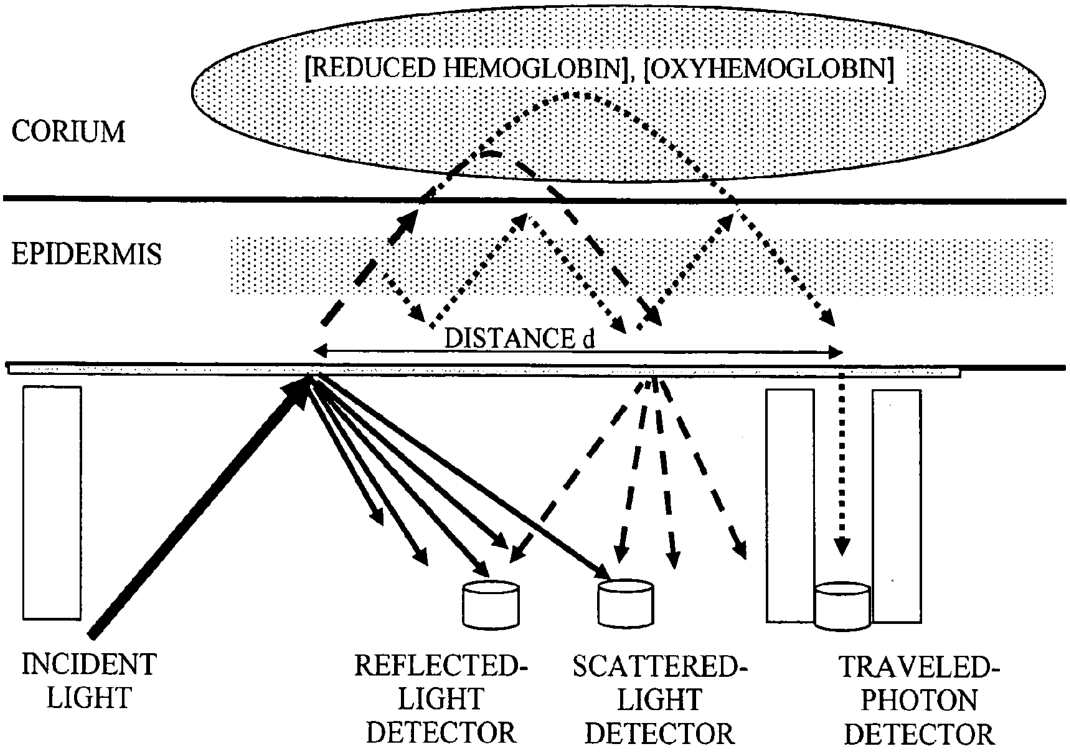


FIG. 1

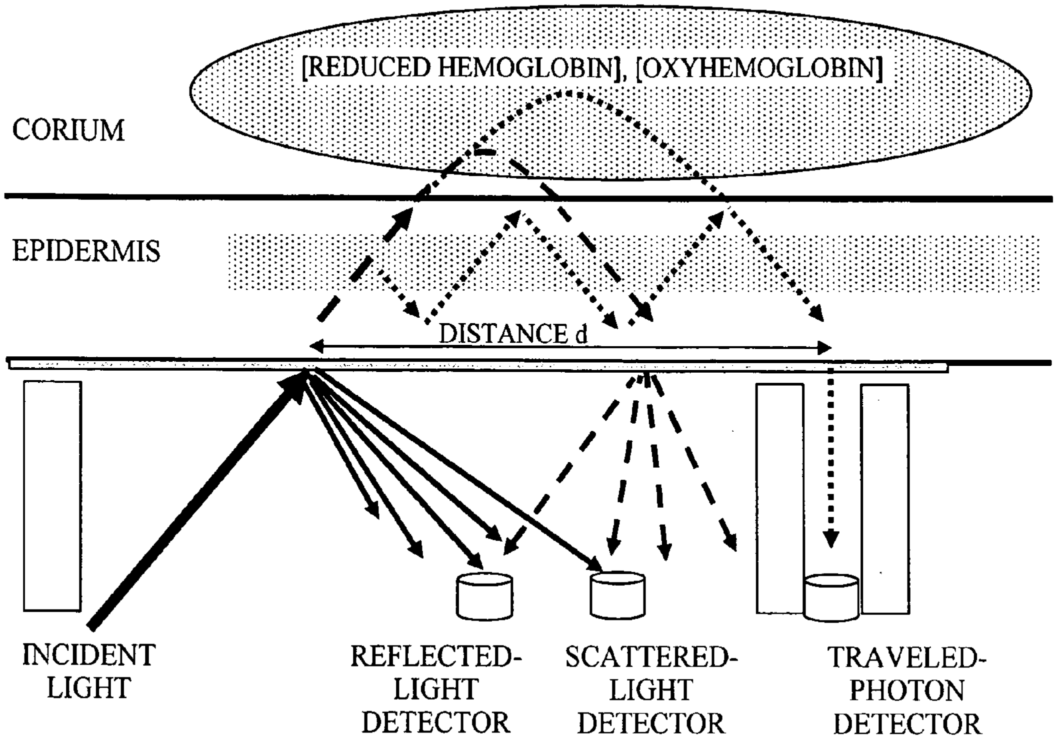

[REDUCED HEMOGLOBIN], [OXYHEMOGLOBIN]
CORIUM
EPIDERMIS
DISTANCE d
INCIDENT LIGHT
REFLECTED-LIGHT DETECTOR
SCATTERED-LIGHT DETECTOR
TRAVELED-PHOTON DETECTOR

FIG. 2

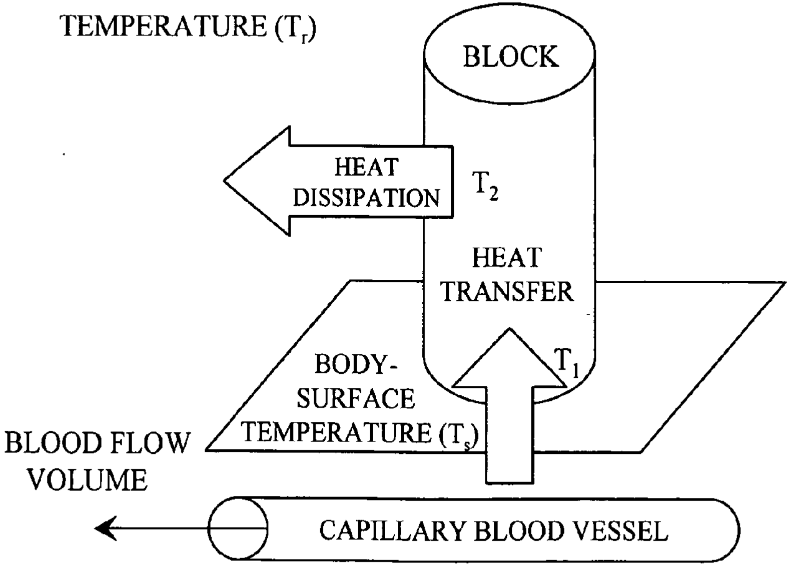

ROOM TEMPERATURE (T_r)
BLOCK
HEAT DISSIPATION
T_2
HEAT TRANSFER
T_1
BODY-SURFACE TEMPERATURE (T_s)
BLOOD FLOW VOLUME
CAPILLARY BLOOD VESSEL

FIG. 3

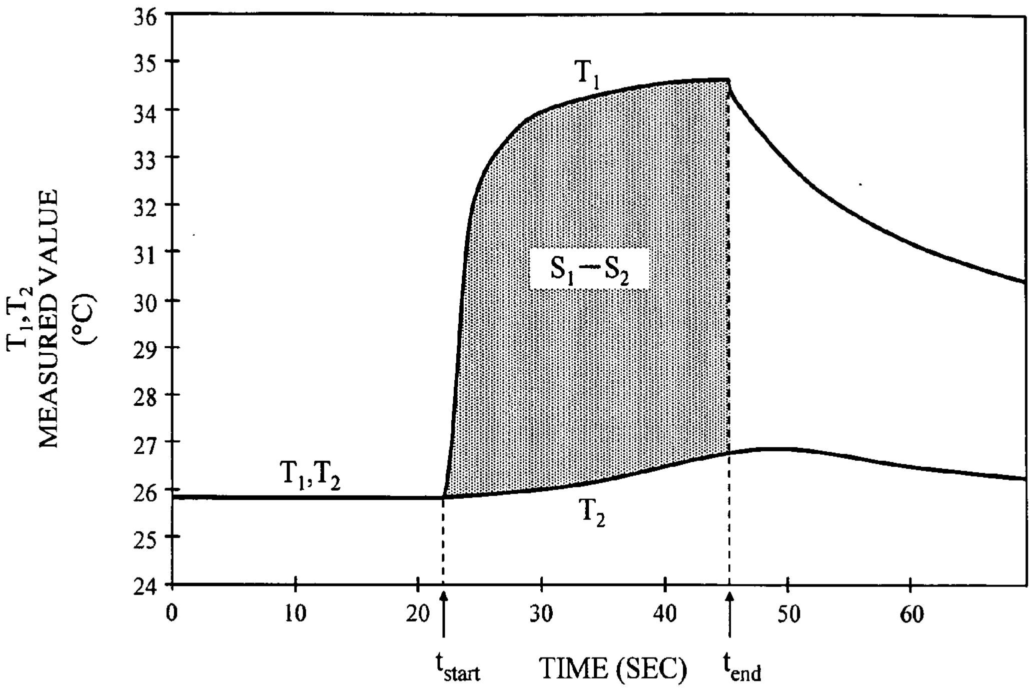

T1,T2
MEASURED VALUE
(°C)
36
35
34
33
32
31
30
29
28
27
26
25
24
T1
S1−S2
T1,T2
T2
0
10
20
30
40
50
60
tstart
TIME (SEC)
tend

FIG. 4

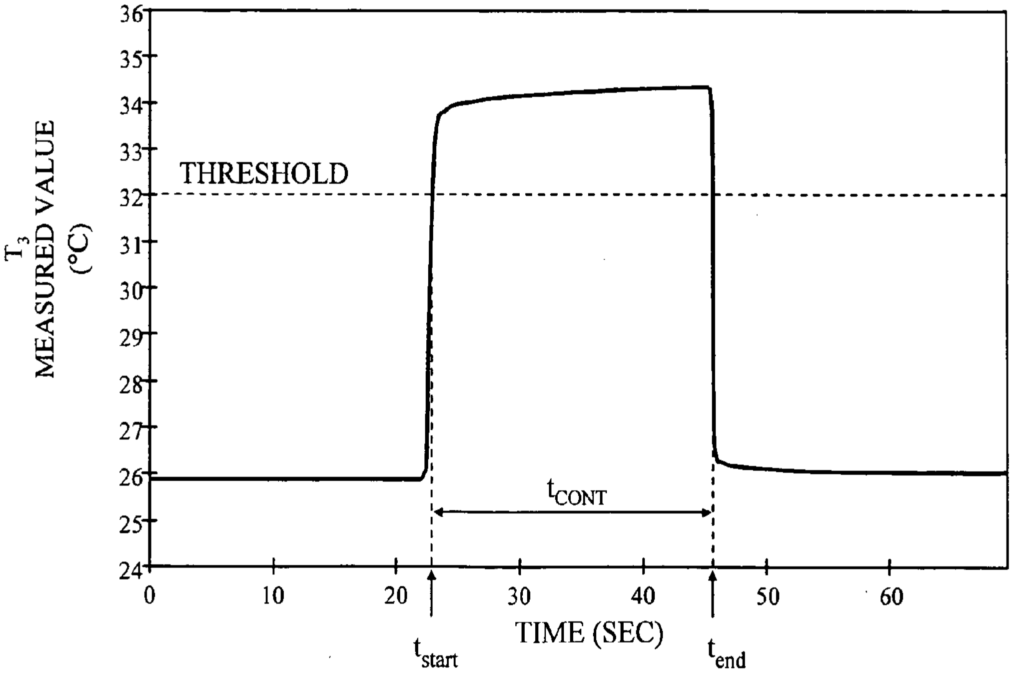

T3
MEASURED VALUE
(°C)
36
35
34
33
32
31
30
29
28
27
26
25
24
THRESHOLD
tCONT
0
10
20
30
40
50
60
tstart
TIME (SEC)
tend

FIG. 5

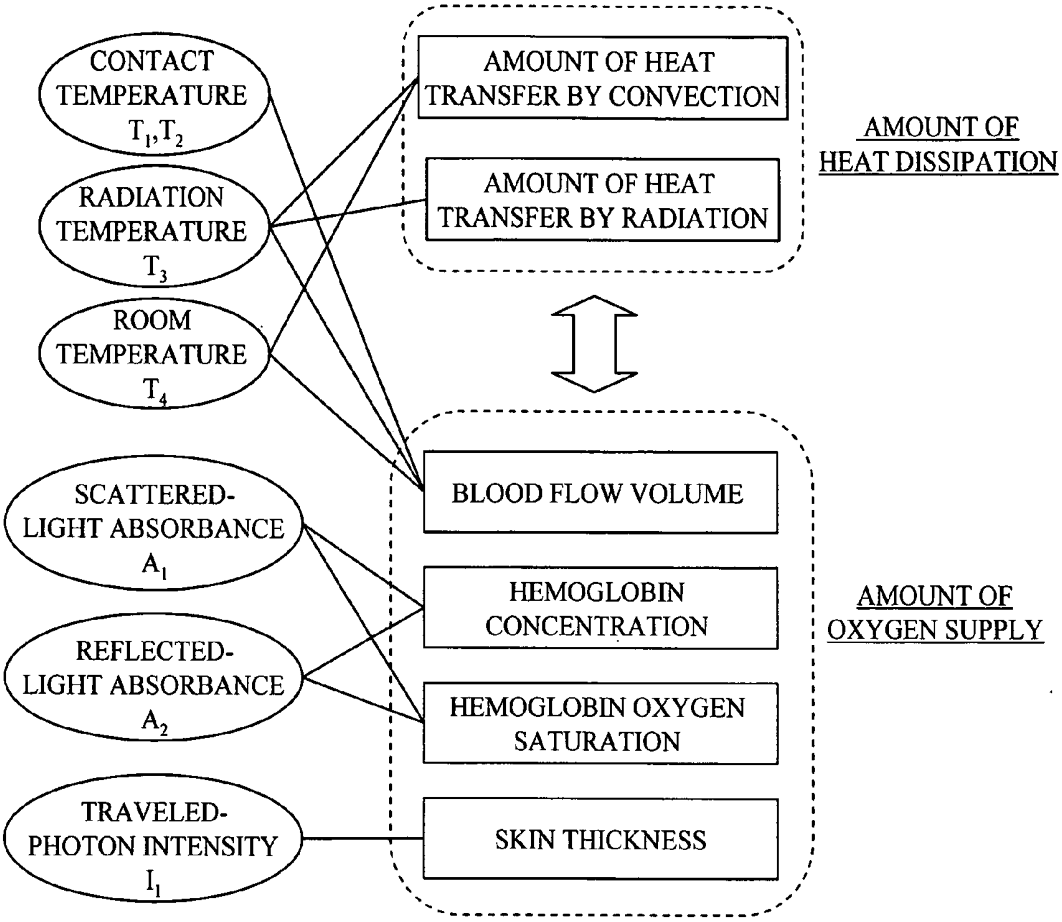

CONTACT TEMPERATURE T_1,T_2
RADIATION TEMPERATURE T_3
ROOM TEMPERATURE T_4
SCATTERED-LIGHT ABSORBANCE A_1
REFLECTED-LIGHT ABSORBANCE A_2
TRAVELED-PHOTON INTENSITY I_1
AMOUNT OF HEAT TRANSFER BY CONVECTION
AMOUNT OF HEAT TRANSFER BY RADIATION
AMOUNT OF HEAT DISSIPATION
BLOOD FLOW VOLUME
HEMOGLOBIN CONCENTRATION
HEMOGLOBIN OXYGEN SATURATION
SKIN THICKNESS
AMOUNT OF OXYGEN SUPPLY

FIG. 6

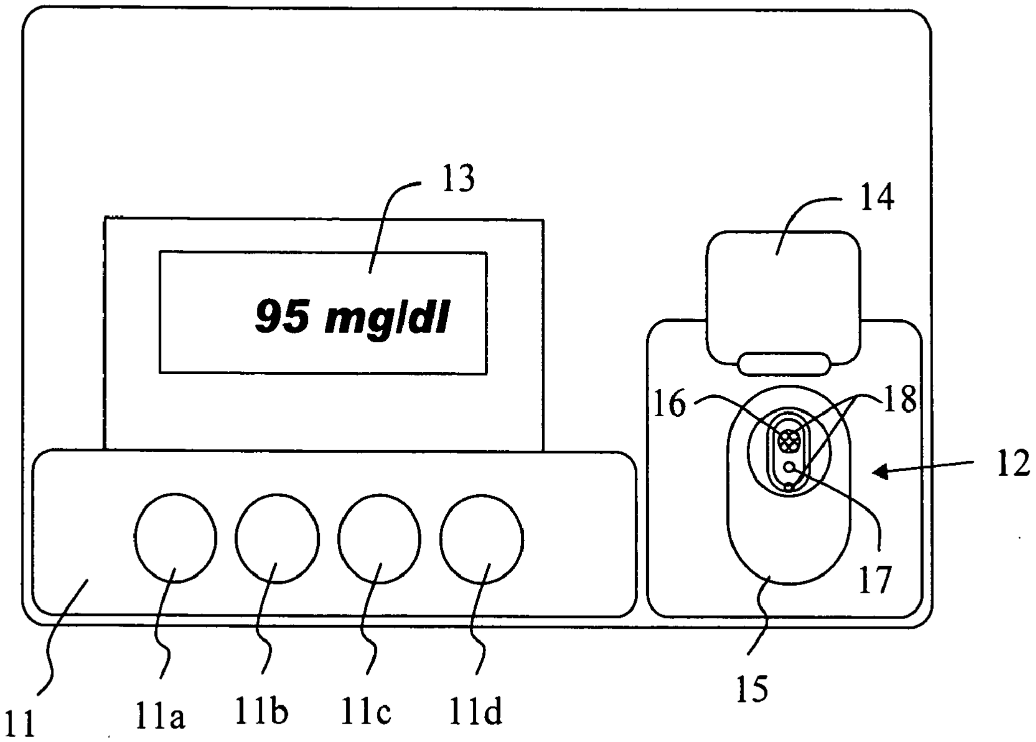


FIG. 7

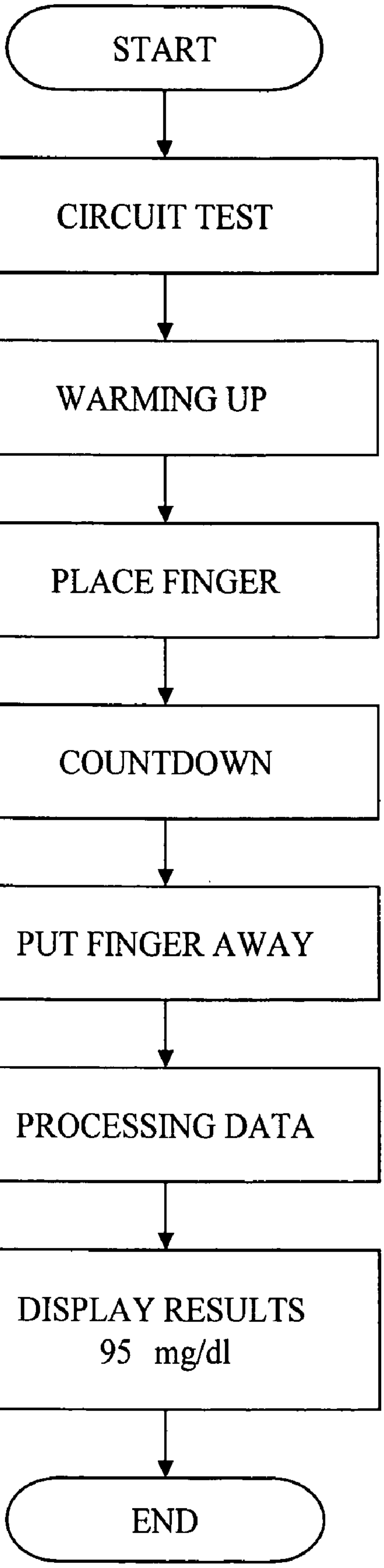

START
CIRCUIT TEST
WARMING UP
PLACE FINGER
COUNTDOWN
PUT FINGER AWAY
PROCESSING DATA
DISPLAY RESULTS
95 mg/dl
END

FIG. 8A

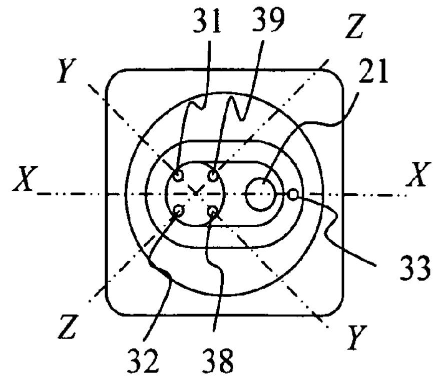


FIG. 8B

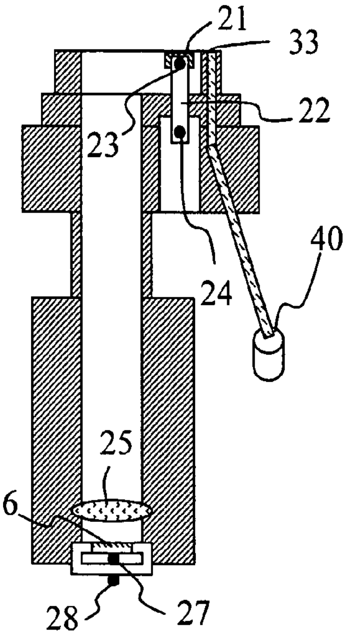


FIG. 8C

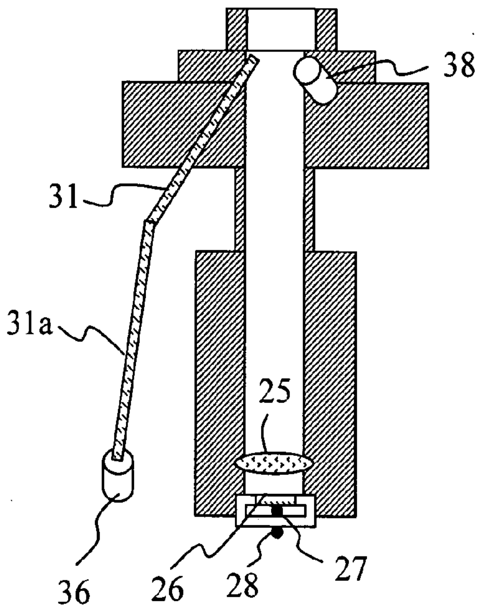


FIG. 8D

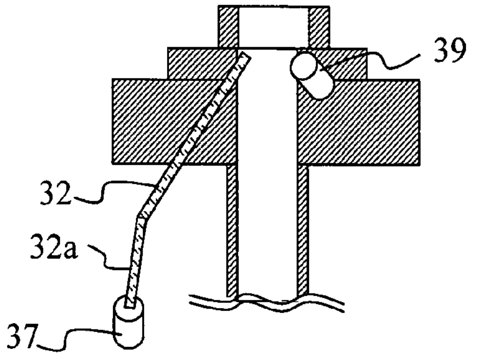


FIG. 8E

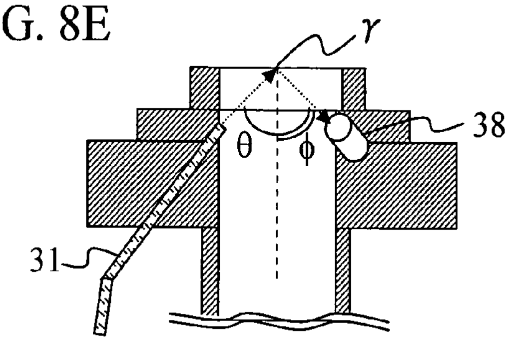


FIG. 9

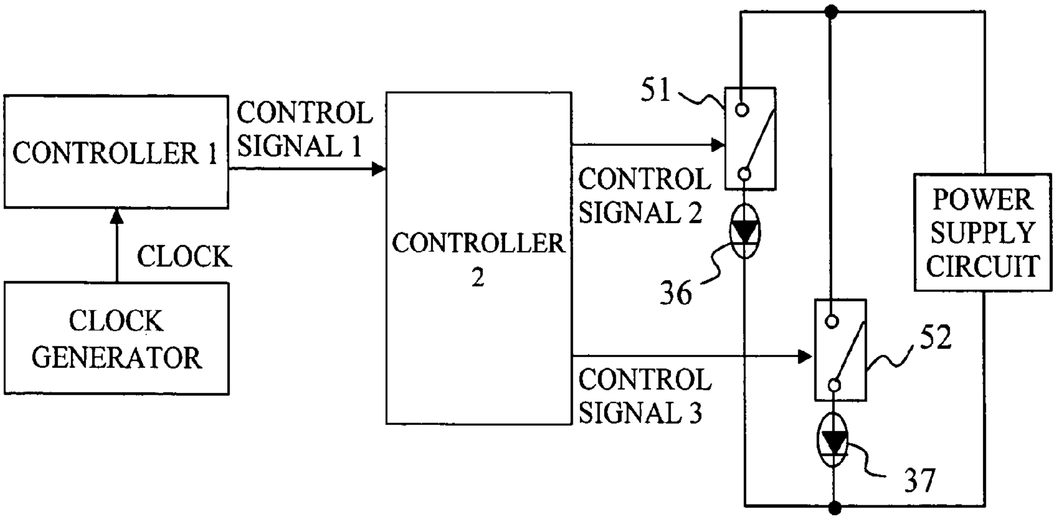

CONTROLLER 1
CONTROL SIGNAL 1
CLOCK
CLOCK GENERATOR
CONTROLLER 2
CONTROL SIGNAL 2
CONTROL SIGNAL 3
51
36
52
37
POWER SUPPLY CIRCUIT

FIG. 10A

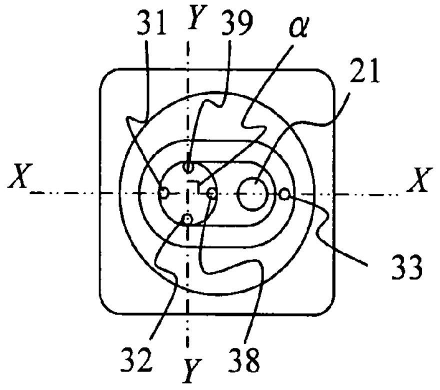


FIG. 10B

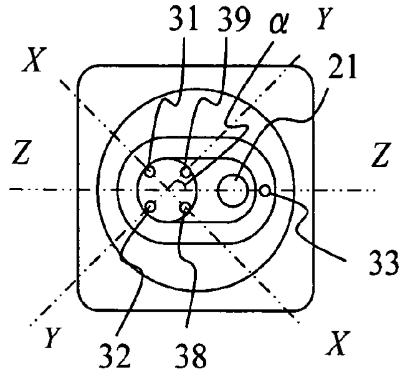


FIG. 10C

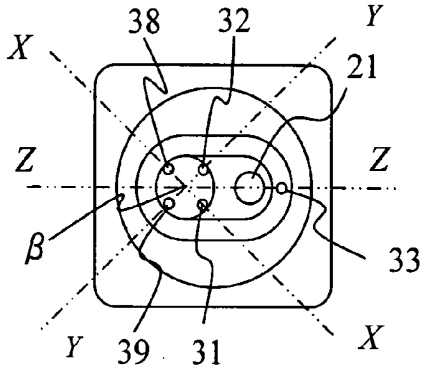


FIG. 10D

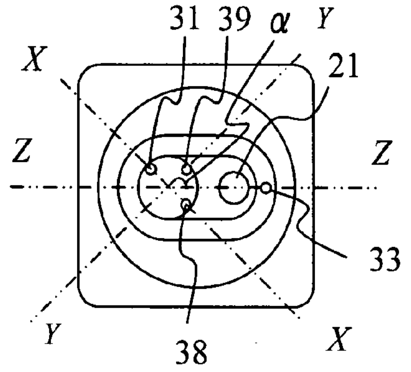


FIG. 10E

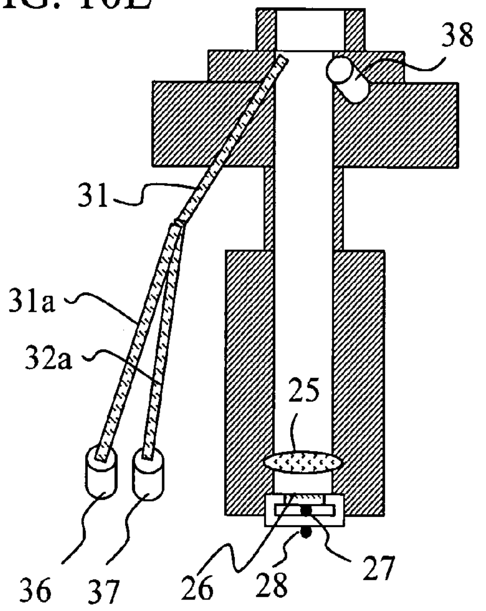


FIG. 10F

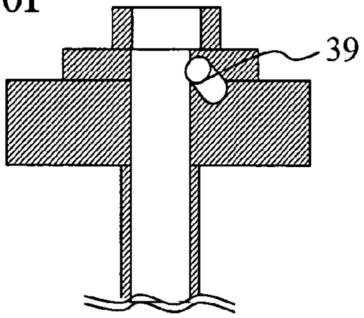


FIG. 10G

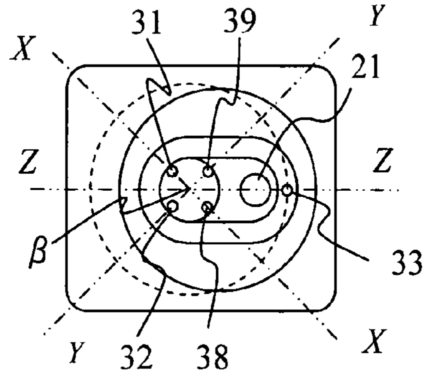


FIG. 11A

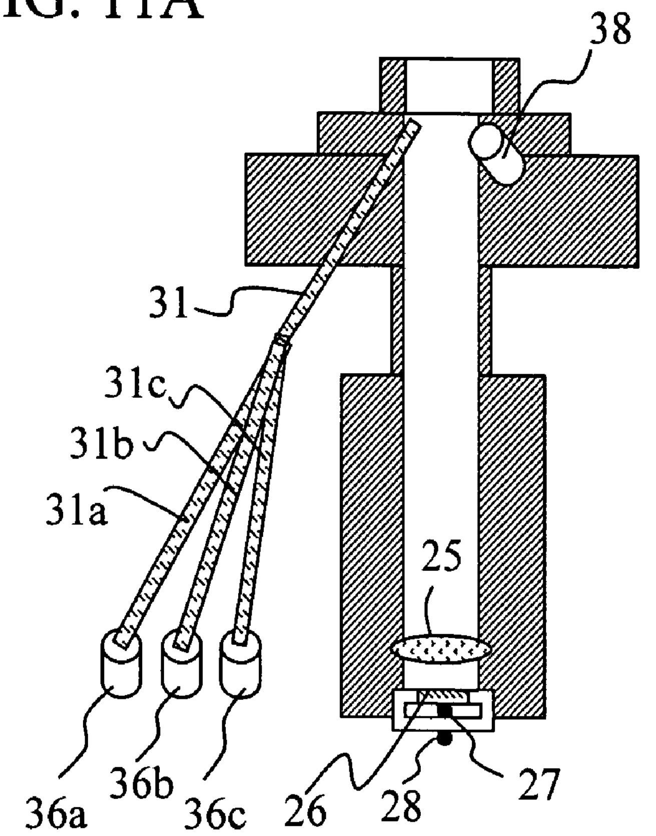


FIG. 11B

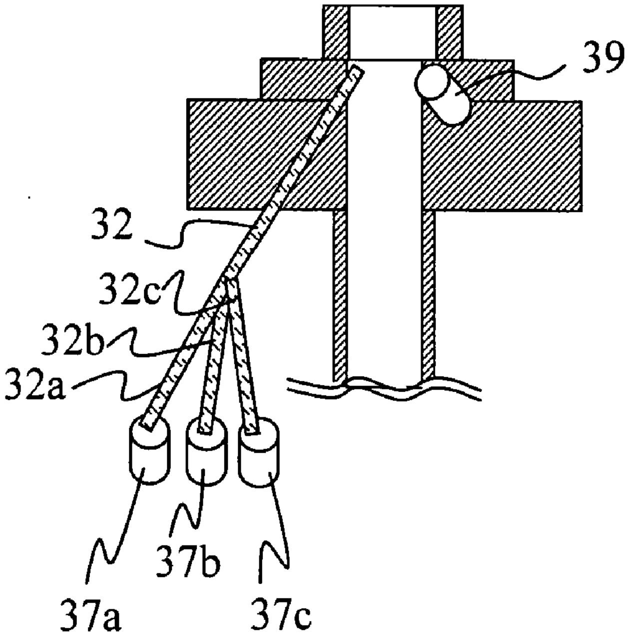


FIG. 12A

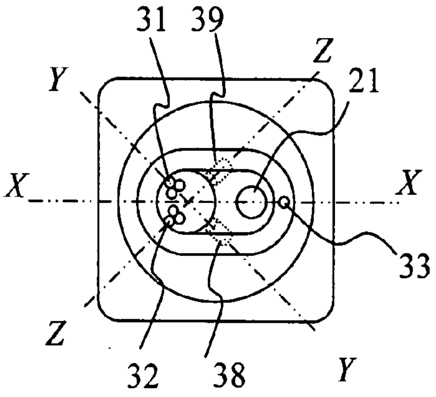


FIG. 12B

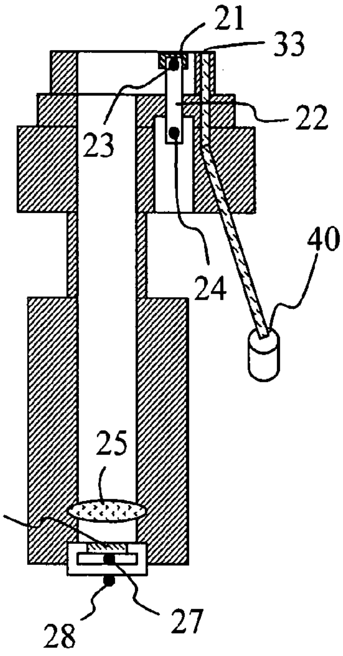


FIG. 12C

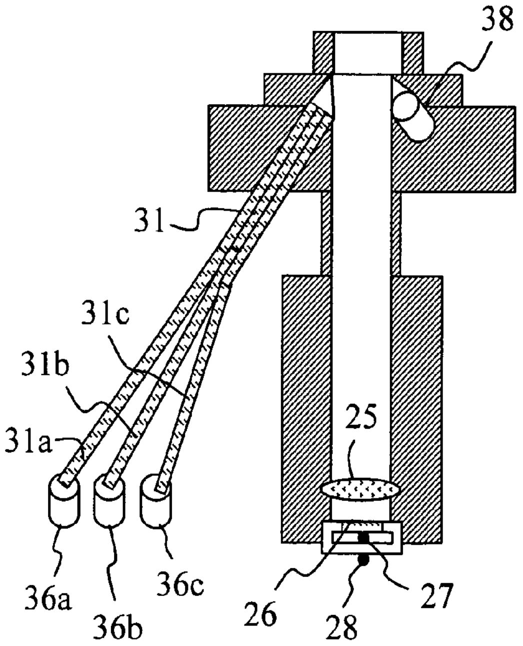


FIG. 12D

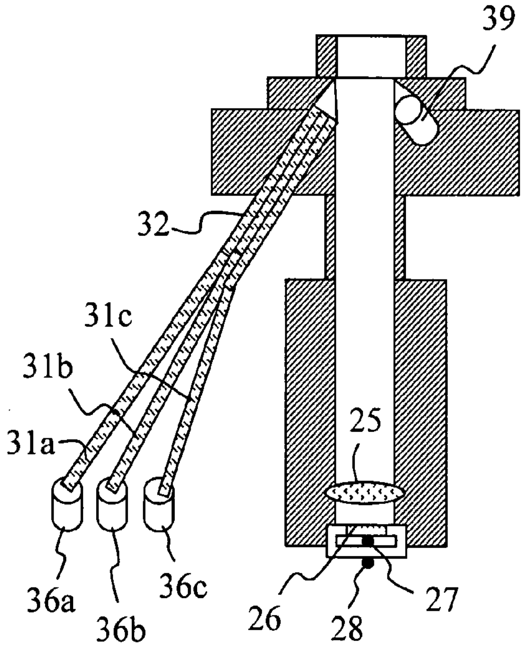


FIG. 13

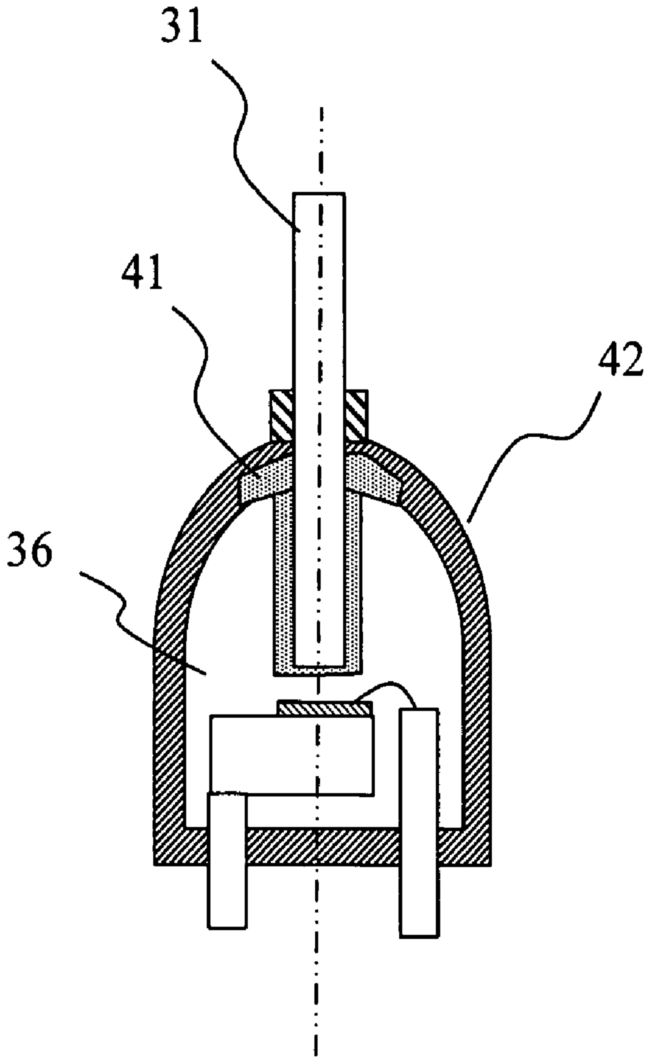


FIG. 14

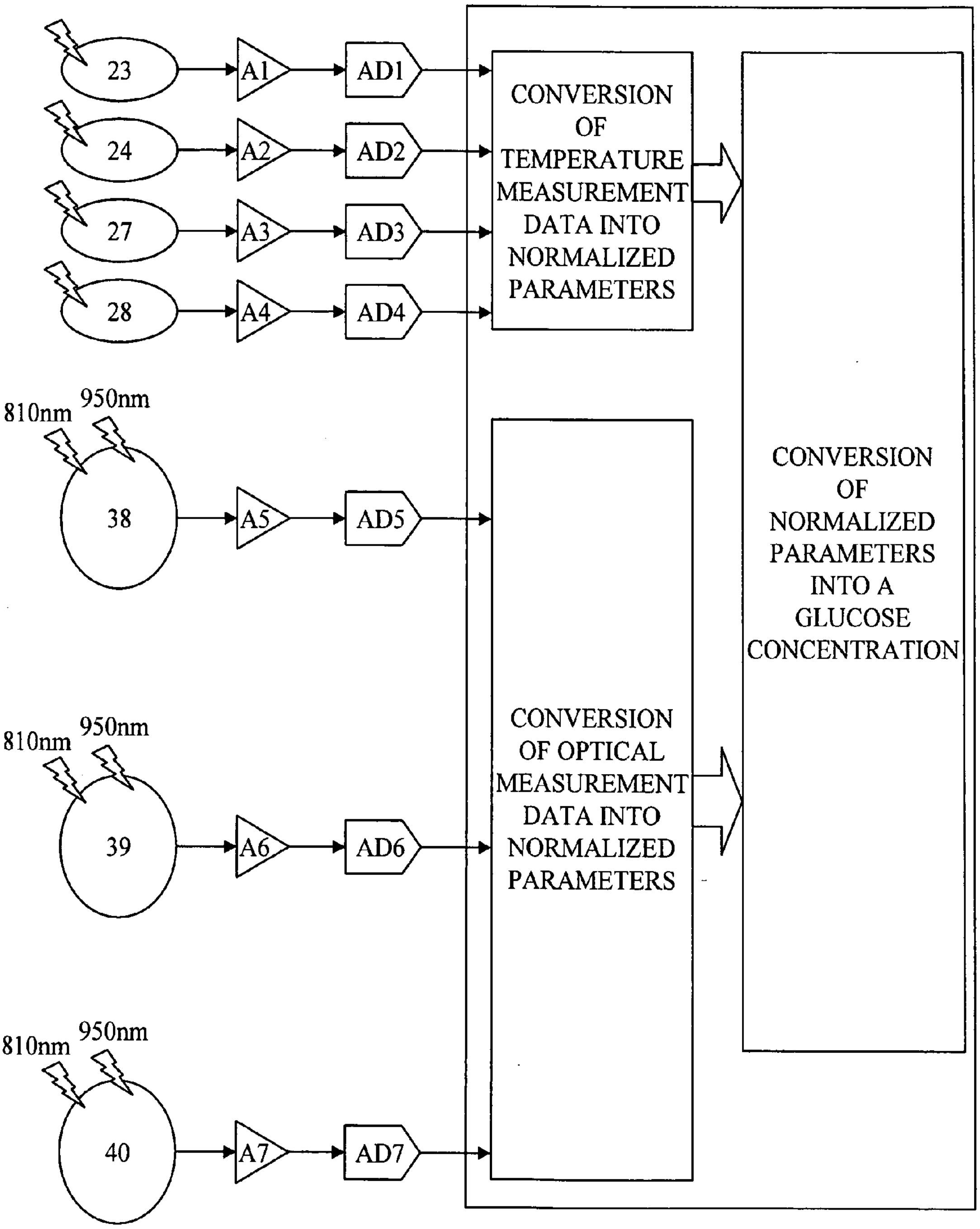
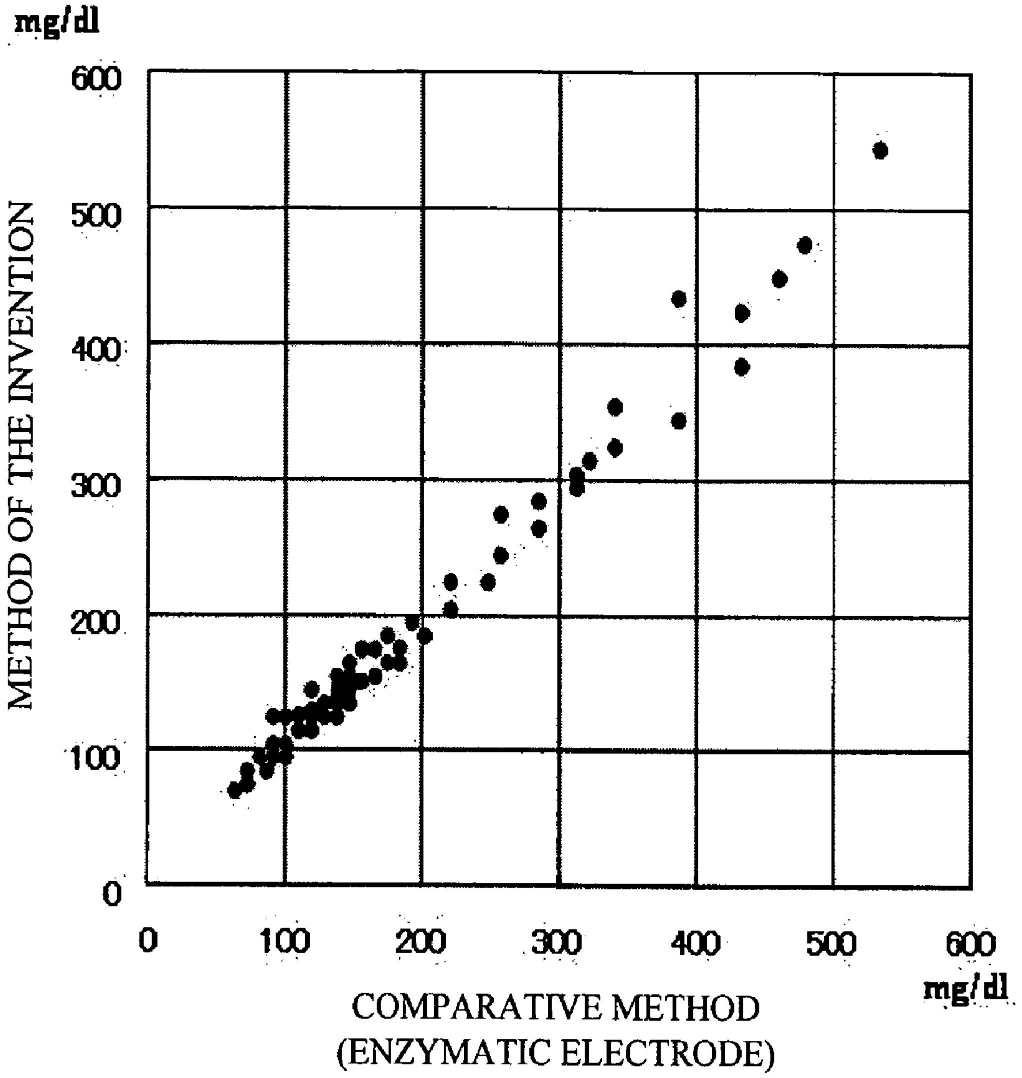

23
24
27
28
A1
A2
A3
A4
AD1
AD2
AD3
AD4
CONVERSION OF TEMPERATURE MEASUREMENT DATA INTO NORMALIZED PARAMETERS
810nm
950nm
38
A5
AD5
810nm
950nm
39
A6
AD6
810nm
950nm
40
A7
AD7
CONVERSION OF OPTICAL MEASUREMENT DATA INTO NORMALIZED PARAMETERS
CONVERSION OF NORMALIZED PARAMETERS INTO A GLUCOSE CONCENTRATION

FIG. 15


mg/dl
600
500
400
300
200
100
0
METHOD OF THE INVENTION
0
100
200
300
400
500
600
mg/dl
COMPARATIVE METHOD
(ENZYMATIC ELECTRODE)

OPTICAL MEASUREMENT APPARATUS AND BLOOD SUGAR LEVEL MEASURING APPARATUS USING THE SAME

CO-PENDING APPLICATION

[0001] U.S. patent application Ser. No. 10/620,689 is a co-pending application of this application. The disclosures of the co-pending application are incorporated herein by cross-reference.

BACKGROUND OF THE INVENTION

[0002] 1. Field of the Invention

[0003] The present invention relates to non-invasive measurement of blood sugar levels for measuring glucose concentration in a living body without blood sampling, and an optical measurement apparatus suitable therefor.

[0004] 2. Description of Related Art

[0005] Hilson et al. report facial and sublingual temperature changes in diabetics following intravenous glucose injection (Diabete & Metabolisme, "Facial and sublingual temperature changes following intravenous glucose injection in diabetics" by R. M. Hilson and T. D. R. Hockaday, 1982, 8, 15-19). Scott et al. discuss the issue of diabetics and thermoregulation (Can. J. Physiol. Pharmacol., "Diabetes mellitus and thermoregulation", by A. R. Scott, T. Bennett, I. A. MacDonald, 1987, 65, 1365-1376). Based on the knowledge gained from such researches, Cho et al. suggest a method and apparatus for determining blood glucose concentration by temperature measurement without requiring the collection of a blood sample (U.S. Pat. Nos. 5,924,996, and 5,795,305).

[0006] Various other attempts have been made to determine glucose concentration without blood sampling. For example, a method has been suggested (JP Patent Publication (Kokai) No. 2000-258343 A) whereby a measurement site is irradiated with near-infrared light of three wavelengths, and the intensity of transmitted light as well as the temperature of the living body is detected. A representative value of the second-order differentiated value of absorbance is then calculated, and the representative value is corrected in accordance with the difference between the living body temperature and a predetermined reference temperature. The blood sugar concentration corresponding to the thus corrected representative value is then determined. An apparatus is also provided (JP Patent Publication (Kokai) No. 10-33512 A (1998)) whereby a measurement site is heated or cooled while monitoring the living body temperature. The degree of attenuation of light based on light irradiation is measured at the moment of temperature change so that the glucose concentration responsible for the temperature-dependency of the degree of light attenuation can be measured. Further, an apparatus is reported (JP Patent Publication (Kokai) No. 10-108857 A (1998)) whereby an output ratio between reference light and transmitted light following the irradiation of the sample is taken, and then a glucose concentration is calculated in accordance with a linear expression of the logarithm of the output ratio and the living body temperature. Another apparatus for measuring glucose concentration is provided (U.S. Pat. No. 5,601,079) whereby the result of irradiation using two light sources is detected by three infrared light detectors and also temperature is detected.

SUMMARY OF THE INVENTION

[0007] Glucose (blood sugar) in blood is used for glucose oxidation reaction in cells to produce necessary energy for the maintenance of living bodies. In the basal metabolism state, in particular, most of the produced energy is converted into heat energy for the maintenance of body temperature. Thus, it can be expected that there is some relationship between blood glucose concentration and body temperature. However, as is evident from the way sicknesses cause fever, the body temperature also fluctuates due to factors other than blood glucose concentration. While methods have been proposed to determine blood glucose concentration by temperature measurement without blood sampling, they could hardly be considered sufficiently accurate.

[0008] Further, a method has also been proposed that detects the result of irradiation of light from two light sources using three infrared light detectors, and that also detects temperature for determining glucose concentration. The method, which only detects two kinds of optical intensity, is unable to provide sufficient accuracy.

[0009] It is an object of the invention to provide a method and apparatus for determining blood glucose concentration with high accuracy based on temperature data and optical data of a test subject without blood sampling.

[0010] Blood sugar is delivered to the cells throughout the human body via blood vessel systems, particularly the capillary blood vessels. In the human body, complex metabolic pathways exist. Glucose oxidation is a reaction in which, fundamentally, blood sugar reacts with oxygen to produce water, carbon dioxide, and energy. Oxygen herein refers to the oxygen delivered to the cells via blood. The volume of oxygen supply is determined by the blood hemoglobin concentration, the hemoglobin oxygen saturation, and the volume of blood flow. On the other hand, the heat produced in the body by glucose oxidation is dissipated from the body by convection, heat radiation, conduction, and so on. On the assumption that the body temperature is determined by the balance between the amount of energy produced in the body by glucose burning, namely heat production, and heat dissipation such as mentioned above, the inventors set up the following model:

(1) The amount of heat production and the amount of heat dissipation are considered equal.

(2) The amount of heat production is a function of the blood glucose concentration and the volume of oxygen supply.

(3) The volume of oxygen supply is determined by the blood hemoglobin concentration, the blood hemoglobin oxygen saturation, and the volume of blood flow in the capillary blood vessels.

(4) The amount of heat dissipation is mainly determined by heat convection and heat radiation.

[0011] According to this model, we achieved the present invention after realizing that blood sugar levels can be accurately determined on the basis of the results of measuring the temperature of the body surface and parameters relating to the blood oxygen concentration and the blood flow volume. The parameters can be measured, e.g., from a part of the human body, such as the fingertip. The parameters relating to convection and radiation can be determined by measuring the temperature on the fingertip. The parameters relating to the blood hemoglobin concentration and the blood hemoglobin oxygen saturation can be determined by spectroscopically measuring blood hemoglobin and then finding the ratio between hemoglobin bound with oxygen and hemoglobin not bound with oxygen. The parameter relating to the volume of blood flow can be determined by measuring the amount of heat transfer from the skin.

[0012] The invention provides an optical measurement apparatus comprising: a first light source for producing light of a first wavelength; a first optical fiber for irradiating a light incident point on the surface of a subject with the light from said first light source; a second light source for producing light of a second wavelength; a second optical fiber for irradiating said light incident point on the surface of the subject with the light from said second light source in a direction different from that of the light of said first wavelength; a first photodetector on which reflected light of the light of said first wavelength reflected by said light incident point and scattered light of the light of said second wavelength are incident without via fiber; a second photodetector on which reflected light of the light of said second wavelength reflected at said light incident point and scattered light of the light of said first wavelength are incident without via fiber; a third photodetector; and a third optical fiber having an incident end thereof disposed at such a position as to be in contact with the surface of said subject, said third optical fiber being adapted to receive, on an incident end thereof, light exiting from an area spaced apart from said light incident point on the surface of said subject and then transmit the light to said third detector.

[0013] Preferably, the plane of incidence of the light of the first wavelength and the plane of incidence of the light of the second wavelength are substantially perpendicular to each other with respect to the light incident point on the subject surface. The plane of incidence herein refers to a plane that includes the incident ray and a normal at the incident point on the subject surface. Further, in the present specification, the ray that enters the incident plane after having been irradiated onto the incident point on the subject surface will be referred to as reflected light. The light that leaves in directions other than that of the incident plane from near the incident point will be referred to as scattered light. The scattered light that leaves out of a position on the subject surface that is spaced apart from the incident point will be referred to as traveled photon.

[0014] Preferably, the outgoing light from each light source is irradiated onto the light incident point on the subject surface via an optical fiber. The reflected light and scattered light from the examined subject are directly incident on the photodetector, and the traveled photon is incident on the photodetector via an optical fiber. An exiting end of the light-irradiating optical fiber and an incident end of the optical fiber for detecting reflected or scattered are preferably disposed near the plane of a cone whose apex corresponds to the light incident point on the subject surface. The first wavelength may be a wavelength at which the molar absorption coefficient of oxyhemoglobin is equal to that of reduced hemoglobin, and the second wavelength may be a wavelength for detecting the difference in absorbance between the oxyhemoglobin and reduced hemoglobin.

[0015] The invention further provides a blood sugar level measuring apparatus comprising: (1) a heat-amount measuring portion for measuring a plurality of temperatures deriving from a body surface and acquiring information that is used for calculating a convective heat transfer amount and radiation heat transfer amount that are related to the dissipation of heat from the body surface; (2) a blood flow volume measuring portion for acquiring information about the volume of blood flow; (3) an optical measurement portion including a light source for producing light of at least two different wavelengths, an optical system for irradiating the body surface with the light emitted by said light source, and at least three different photodetectors for detecting the light that has been irradiated onto the body surface, said optical measurement portion providing hemoglobin concentration and hemoglobin oxygen saturation in blood; (4) a memory portion in which relationships between parameters respectively corresponding to said plurality of temperatures, blood flow volume, hemoglobin concentration and hemoglobin oxygen saturation in blood, and blood sugar levels are stored; (5) a calculation portion for converting a plurality of measurement values inputted from said heat amount measuring portion, said blood flow volume measuring portion, and said optical measurement portion respectively into said parameters, and then calculating a blood sugar value by applying said parameters to said relationships stored in said memory portion; and (6) a display portion for displaying the blood sugar level calculated by said calculation portion, wherein said optical measurement portion includes a first light source for producing light of a first wavelength, a second light source for producing light of a second wavelength, a first optical fiber, a second optical fiber, a third optical fiber, a first photodetector, a second photodetector, and a third photodetector, a light incident point on the surface of a subject is irradiated with light emitted by said first light source via said first optical fiber, said light incident point on the surface of said subject is irradiated with light emitted by said second light source via said second optical fiber in a direction different from that of the light of said first wavelength, reflected light of the light of said first wavelength reflected at said light incident point and scattered light of the light of said second wavelength are incident on said first photodetector without via fiber, reflected light of the light of said second wavelength reflected at said light incident point and scattered light of the light of said first wavelength are incident on said second photodetector without via fiber, said third optical fiber has an incident end thereof disposed at such a position as to be in contact with the surface of said subject in an area spaced apart from said light incident point on the subject surface, and said third photodetector is adapted to receive light exiting from an area spaced apart from said light incident point on the surface of the subject via said third optical fiber.

BRIEF DESCRIPTION OF THE DRAWINGS

[0016] **FIG. 1** shows a model of the transmission of light in the case of irradiating the skin surface with continuous light.

[0017] **FIG. 2** shows a model of heat transfer from the body surface to a block.

[0018] **FIG. 3** shows a temporal change in measurement values of temperatures T_1 and T_2.

[0019] **FIG. 4** shows an example of a measurement of a temporal change in temperature T_3.

[0020] **FIG. 5** shows the relationships between measurement values provided by various sensors and the parameters derived therefrom.

[0021] **FIG. 6** shows an upper plan view of a non-invasive blood sugar level measuring apparatus according to the present invention.

[0022] **FIG. 7** shows the operating procedure for the apparatus.

[0023] **FIGS. 8A** to **8**E show a measuring portion in detail.

[0024] **FIG. 9** shows a block diagram of an example of a circuit for causing light-emitting diodes to emit light in a time-divided manner.

[0025] **FIGS. 10A** to **10**G show an optical sensor portion and the measuring portion in detail.

[0026] **FIGS. 11A** to **11**B show in detail the measuring portion for a plurality of wavelengths.

[0027] **FIGS. 12A** to **12**D show in detail the measuring portion for a plurality of wavelengths.

[0028] **FIG. 13** shows the connection and blocking of light between a light-emitting diode and an optical fiber.

[0029] **FIG. 14** shows a conceptual chart illustrating the flow of data processing in the apparatus.

[0030] **FIG. 15** shows a chart plotting the glucose concentration values calculated according to the present invention and the glucose concentration values measured by the enzymatic electrode method.

DETAILED DESCRIPTION OF THE PREFERRED EMBODIMENTS

[0031] The invention will now be described by way of preferred embodiments thereof with reference made to the drawings.

[0032] Initially, the above-mentioned model will be described in more specific terms. Regarding the amount of heat dissipation, convective heat transfer, which is one of the main causes of heat dissipation, is related to temperature difference between the ambient (room) temperature and the body-surface temperature. The amount of heat dissipation due to radiation, which is another main cause of dissipation, is proportional to the fourth power of the body-surface temperature according to the Stefan-Boltzmann law. Thus, it can be seen that the amount of heat dissipation from the human body is related to the room temperature and the body-surface temperature. On the other hand, the amount of oxygen supply, which is a major factor related to the amount of heat production, is expressed as the product of hemoglobin concentration, hemoglobin oxygen saturation, and blood flow volume.

[0033] The hemoglobin concentration can be measured from the absorbance at the wavelength (equal-absorbance wavelength) at which the molar absorbance coefficient of the oxyhemoglobin is equal to that of the reduced (deoxy-) hemoglobin. The hemoglobin oxygen saturation can be measured by measuring the absorbance at the equal-absorbance wavelength and the absorbance at at least one different wavelength at which the ratio between the molar absorbance coefficient of the oxyhemoglobin and that of the reduced (deoxy-) hemoglobin is known, and then solving simultaneous equations. Namely, the hemoglobin concentration and hemoglobin oxygen saturation can be obtained by conducting the measurement of absorbance at at least two wavelengths. However, in order to accurately determine the hemoglobin concentration and hemoglobin oxidation saturation from absorbance, the influence of interfering components must be corrected. The interfering components affecting the absorbance include the thickness of the skin (epidermis), for example. These interfering components can be measured in various manners, of which one example will be described below.

[0034] The thickness of the skin can be measured by measuring the intensity of only that light that has traveled in the skin by a distance d from where light was shone on the skin. **FIG. 1** shows the behavior of light in the case where the skin surface was irradiated with continuous light. As the light of a certain wavelength and intensity is shone, the light is reflected and scattered by the skin surface. Part of the light penetrates the skin and experiences scattering and diffusion in a repeated manner. In such a behavior of light, the depth of penetration of the light that has traveled by distance d is substantially constant depending on the wavelength. The skin does not contain blood, so it has a low fluidity, resulting in a low absorbance. On the other hand, the corium contains blood and therefore has a high fluidity, resulting in a high absorbance. Thus, when the skin is thin, the light can penetrate deeper into the corium, resulting in a larger absorbance. When the skin is thick, the distance traveled by the light becomes shorter, so that the absorbance becomes smaller. By taking the ratio between the detected intensity of only that light that has traveled distance d and the detected intensity of the light that has traveled in a standard substance with a known thickness in the same manner, the thickness of the skin can be estimated.

[0035] The measurements are carried out using at least three detectors, namely a reflected-light detector for detecting mainly reflected light, a scattered-light detector for detecting mainly scattered light, and a traveled-photon detector for detecting traveled photon.

[0036] The reflected-light detector can detect part of the scattered light produced by the light traveling inside the body and then exiting from the body surface, as well as mainly the reflected light reflected by the body surface. The scattered-light detector can detect part of the scattered light scattered on the body surface, as well as mainly the scattered light produced by the light passing inside the body and then exiting through the body surface. The path of the traveled photon up to the traveled-photon detector is optically blocked in order to prevent the detection of light other than the traveled photon, namely the light deriving from reflected light and scattered light. The traveled-photon detector is thus adapted to detect only traveled photon, so that the skin thickness can be estimated. During detection, a total of at least three detectors, namely at least one each of the reflected-light detector, scattered-light detector, and traveled-photon detector, are used. Preferably, additional detectors with similar functions and high detection sensitivities adapted for particular kinds of wavelength may be used. Further, a transmitted-light detector may be added for detecting light that has passed through the detection area, as necessary.

[0037] The wavelength values described herein are most appropriate values for obtaining absorbance for various intended purposes, such as for obtaining the absorbance at the equal molar absorbance coefficient wavelengths, or for obtaining the peak of absorbance. Thus, in addition to the wavelengths described herein, other wavelengths in the vicinities thereof may be used and still similar measurements can be performed.

[0038] The rest is the blood flow volume, which can be measured by various methods. One example will be described below.

[0039] **FIG. 2** shows a model for the description of the transfer of heat from the body surface to a solid block having a certain heat capacity when the block is brought into contact with the body surface for a certain time and then separated. The block is made of resin such as plastic or vinyl chloride. In the illustrated example, attention will be focused on the temporal variation of the temperature T_1 of a portion of the block that is brought into contact with the body surface, and the temporal variation of the temperature T_2 at a point on the block spaced apart from the body surface. The blood flow volume can be estimated by monitoring mainly the temporal variation of the temperature T_2 (at the spatially separated point on the block). The details will follow.

[0040] Before the block comes into contact with the body surface, the temperatures T_1 and T_2 at the two points of the block are equal to the room temperature T_r. When a body-surface temperature T_s is higher than the room temperature T_r as the block comes into contact with the body surface, the temperature T_1 swiftly rises due to the transfer of heat from the skin, and it approaches the body-surface temperature T_s. On the other hand, the temperature T_2 is lowered from the temperature T_1 as the heat conducted through the block is dissipated from the block surface, and it rises more gradually. The temporal variation of the temperatures T_1 and T_2 depends on the amount of heat transferred from the body surface to the block, which in turn depends on the blood flow volume in the capillary blood vessels under the skin. If the capillary blood vessels are regarded as a heat exchanger, the coefficient of transfer of heat from the capillary blood vessels to the surrounding cell tissues is given as a function of the blood flow volume. Thus, by measuring the amount of heat transfer from the body surface to the block by monitoring the temporal variation of the temperatures T_1 and T_2, the amount of heat transferred from the capillary blood vessels to the cell tissues can be estimated. Based on this estimation, the blood flow volume can then be estimated.

[0041] **FIG. 3** shows the temporal variation of the measured values of the temperature T_1 at the portion of the block in contact with the body surface and the temperature T_2 at the position on the block spaced apart from the body-surface contact position. As the block comes into contact with the body surface, the T_1 measured value swiftly rises, and it gradually drops as the block is brought out of contact.

[0042] **FIG. 4** shows the temporal variation of the value of the temperature T_3 measured by a radiation-temperature detector. As the detector detects the temperature T_3 that is due to radiation from the body surface, it is more sensitive to temperature changes than other sensors. Because radiation heat propagates as an electromagnetic wave, it can transmit temperature changes instantaneously. Thus, by locating the radiation-temperature detector near where the block contacts the body surface so as to detect radiated heat from the body surface, as shown in **FIG. 8** (which will be described later), the time of start of contact t_{start} and the time of end of contact tend between the block and the body surface can be detected from changes in the temperature T_3. For example, a temperature threshold value is set as shown in **FIG. 4**. The contact start time t_{start} is when the temperature threshold value is exceeded. The contact end time tend is when the temperature T_3 drops below the threshold. The temperature threshold is set at 32° C., for example.

[0043] Then, the T_1 measured value between t_{start} and tend is approximated by an S curve, such as a logistic curve. A logistic curve is expressed by the following equation:

$$T = \frac{b}{1 + c \times \exp(-a \times t)} + d$$

where T is temperature, and t is time.

[0044] The measured value can be approximated by determining coefficients a, b, c, and d using the non-linear least-squares method. For the resultant approximate expression, T is integrated between time t_{start} and time tend to obtain a value S_1.

[0045] Similarly, an integrated value S_2 is calculated from the T_2 measured value. The smaller (S_1-S_2) is, the larger the amount of transfer of heat is from the body surface to the position of T_2. (S_1-S_2) becomes larger with increasing body-surface contact time t_{CONT} $(=t_{end}-t_{start})$. Thus, $a_5/(t_{CONT}\times(S_1-S_2))$ is designated as a parameter X_5 indicating the volume of blood flow, using a_5 as a proportionality coefficient.

[0046] Thus, it will be seen that the measured amounts necessary for the determination of blood glucose concentration by the above-described model are the room temperature (ambient temperature), body surface temperature, temperature changes in the block brought into contact with the body surface, the temperature due to radiation from the body surface, the absorbance of reflected light or scattered light at at least two wavelengths, and the intensity of traveled photon.

[0047] **FIG. 5** shows the relationships between the measured values provided by various sensors and the parameters derived therefrom. A block is brought into contact with the body surface, and chronological changes in two kinds of temperatures T_1 and T_2 are measured by two temperature sensors provided at two locations of the block. Separately, radiation temperature T_3 on the body surface and room temperature T_4 are measured. Absorbance A_1 and A_2 of scattered light and reflected light, respectively, are measured at at least two wavelengths related to the absorption of hemoglobin. The intensity I_1 of traveled photon is measured at at least one wavelength. Alternatively, the intensity may be determined by measuring at the aforementioned two wavelengths, and then finding an averaged or mean value from the results thereof. The temperatures T_1, T_2, T_3, and T_4 provide parameters related to the volume of blood flow. The temperature T_3 provides a parameter related to the amount of heat transferred by radiation. The temperatures T_3 and T_4 provide parameters related to the amount of heat transferred by convection. The absorbance A_1 and A_2 and intensity I_1

provide parameters related to the hemoglobin concentration and the hemoglobin oxygen saturation.

[0048] Hereafter, an example of an apparatus for non-invasively measuring blood sugar levels according to the principle of the invention will be described.

[0049] **FIG. 6** shows a top plan view of a non-invasive blood sugar level measuring apparatus according to the invention. While in this example the skin on the ball of the fingertip is used as the body surface, other parts of the body surface may be used.

[0050] On the top surface of the apparatus are provided an operating portion **1**, a measuring portion **12** where the finger to be measured is to be placed, and a display portion **13** for displaying measurement results, the state of the apparatus, measured values, for example. The operating portion **11** includes four push buttons **11***a* to **11***d* for operating the apparatus. The measuring portion **12** has a cover **14** which, when opened (as shown), reveals a finger rest portion **15** with an oval periphery. The finger rest portion **15** accommodates an opening end **16** of a radiation-temperature sensor portion, a contact-temperature sensor portion **17**, and an optical sensor portion **18**.

[0051] **FIG. 7** shows the procedure for operating the apparatus. As a power button on the operating portion is pressed to turn on the apparatus, an indication "Warming up" is displayed on the LCD and the electronic circuits in the apparatus are warmed up. At the same time, a check program is activated to automatically check the electronic circuits. After the warm-up phase is finished, an indication "Place finger" appears on the LCD. As the user places his or her finger on the finger rest portion, a countdown is displayed on the LCD. When the countdown is over, an indication "Put finger away" appears on the LCD. As the user puts his or her finger away, the LCD indicates "Processing data." Thereafter, the display shows a blood sugar level, which is then stored in an IC card together with the date and time. After the user reads the displayed blood sugar level, he or she pushes a particular button on the operating portion. About one minute later, the apparatus displays a message "Place finger" on the LCD, thus indicating that the apparatus is ready for the next cycle of measurement.

[0052] **FIGS. 8A** to **8**E show the measuring portion in detail. **FIG. 8A** is a top plan view, **FIG. 8B** is a cross section taken along line X-X of **FIG. 8A**, **FIG. 8C** is a cross section taken along line Y-Y of **FIG. 8A**, and **FIG. 8D** is a cross section taken along Z-Z of **FIG. 8A**.

[0053] First, the process of measuring temperatures by the non-invasive blood sugar level measuring apparatus according to the invention will be described. In a portion of the measuring portion with which the examined portion (ball of the finger) is to come into contact, a thin plate **21** of a highly heat-conductive material, such as gold, is placed. A bar-shaped heat-conductive member **22**, which is made of a material with a heat conductivity lower than that of the plate **21**, such as polyvinylchloride, is thermally connected to the plate **21** and extends into the apparatus. The temperature sensors include a thermistor **23** that is an adjacent-temperature detector with respect to the examined portion for measuring the temperature of the plate **21**, and a thermistor **24** that is an indirect-temperature detector with respect to the examined portion for measuring the temperature of a portion of the heat-conducting member which is spaced apart from the plate **21** by a certain distance. An infrared lens **25** is disposed inside the apparatus at such a position that the examined portion (ball of the finger) placed on the finger rest portion **15** can be seen through the lens. Below the infrared lens **25** is disposed a pyroelectric detector **27** via an infrared radiation-transmitting window **26**. Another thermistor **28** is disposed in close proximity to the pyroelectric detector **27**.

[0054] Thus, the temperature sensor portion of the measuring portion has four temperature sensors, and they measure four kinds of temperatures as follows:

(1) Temperature on the finger surface (thermistor **23**): T_1

(2) Temperature of the heat-conducting member (thermistor **24**): T_2

(3) Temperature of radiation from the finger (pyroelectric detector **27**): T_3

(4) Room temperature (thermistor **28**): T_4

[0055] The optical sensor portion **18** is described hereafter. The optical sensor portion **18** measures the hemoglobin concentration and the hemoglobin oxygen saturation necessary for the determination of the oxygen supply volume. In order to measure the hemoglobin concentration and the hemoglobin oxygen saturation, it is necessary to measure the absorbance of scattered light at at least two wavelengths, the absorbance of reflected light at at least one wavelength, and the intensity of traveled photon at at least one wavelength. The accuracy of the absorbance of reflected light can be improved by measuring at a plurality of wavelengths, if possible, and then using a mean value. Thus, in the present embodiment, the absorbance of reflected light is measured at two different wavelengths. The accuracy of the measurement of the intensity of traveled photon can also be improved by measuring at a plurality of wavelengths, if possible, and then using a mean value. **FIGS. 8B** to **8**E show exemplary configurations for carrying out the measurement using two light sources **36** and **37** and three detectors **38** to **40**.

[0056] The ends of three optical fibers **31** to **33** are located in the optical sensor portion **18**. The optical fibers **31** and **32** are for optical irradiation, while the optical fiber **33** is for receiving light. As shown in **FIG. 8C**, the optical fiber **31** connects to a branch optical fiber **31***a* that is provided with a light-emitting diode **36** of a single wavelength at the end thereof. Similarly, the optical fiber **32** is connected to a branch optical fiber **32***a* that is provided at the end thereof with a light-emitting diode **37** of a single wavelength. Photodiodes **38** and **39** are disposed in the optical sensor portion. The other end of the light-receiving optical fiber **33** is provided with a photodiode **40**. To each the optical fibers **31** and **32**, there may be connected a plurality of branch optical fibers provided at the ends thereof with light-emitting diodes. The light-emitting diode **36** emits light with a wavelength of 810 nm, while the light-emitting diode **37** emits light with a wavelength of 950 nm. The wavelength 810 nm is the equal absorbance wavelength at which the molar absorbance coefficient of the oxyhemoglobin is equal to that of the reduced (deoxy-) hemoglobin. The wavelength 950 nm is the wavelength at which the difference between the molar absorbance coefficient of the oxyhemoglobin and that of the reduced hemoglobin is large.

[0057] The two light-emitting diodes **36** and **37** emit light in a time-sharing manner. The finger of an examined subject is irradiated with the light emitted by the light-emitting diodes **36** and **37** via the irradiating optical fibers **31** and **32**. The light shone on the finger from the light-irradiating optical fiber **31** is reflected by the skin, and the reflected light is detected by the photodiode **38**, the scattered light is detected by the photodiode **39**, and the traveled photon is incident on the light-receiving optical fiber **33** and is then detected by the photodiode **40**. The traveled photon-receiving optical fiber **33** is adapted to be in close contact with the finger surface such that it can avoid the direct entry of reflected and/or scattered light. The light with which the finger is irradiated via the light-irradiating optical fiber **32** is reflected by the skin of the finger, and the reflected light is detected by the photodiode **39**, the scattered light is detected by the photodiode **38**, and the traveled photon is incident on the light-receiving optical fiber **35** and is then detected by the photodiode **40**. Thus, by causing the two light-emitting diodes **36** and **37** to emit light in a time-divided manner, the photodiodes **38** and **39** can detect different light depending on the irradiating position of the light-irradiating optical fibers. By this structure, the size of the optical sensor portion **18** can be reduced. With regard to the light shone on the finger via the light-irradiating fiber **32**, the light-receiving optical fiber **33** may be adapted not to detect traveled photon.

[0058] By thus employing the structure such that the light reflected by the skin of the finger and the scattered light are received directly by photodiodes, the received amount of light detected by each photodiode can be increased. Regarding the light-receiving optical fiber **33**, it is similarly possible to increase the amount of received light by disposing the photodiode **40** directly at the position corresponding to the end of the light-receiving optical fiber **33**. However, putting the photodiode **40** directly at the end of the light-receiving optical fiber **33** would result in an increased size of the optical sensor portion **18**. Accordingly, it is desirable to use the light-receiving optical fiber **33** if the size of the optical sensor portion **18** is to be further reduced.

[0059] **FIG. 9** shows a block diagram of an example of the circuit for causing the light-emitting diodes to emit in a time-divided manner. A controller **1** causes the light-emitting diodes **36** and **37** to emit light in a time-divided manner by repeating the following steps (1) and (2). **FIG. 9** concerns the case of using two wavelengths (two LEDs).

[0060] (1) Sends a control signal **1** to a controller **2** in synchronism with the clock of a clock generator for a certain duration of time in order to select a control signal **2**. As a result, a switching circuit **51** is turned on, thereby turning power on and causing the light-emitting diode **36** to emit light.

[0061] (2) After a certain duration of time has elapsed, sends a control signal **1** to the controller **2** in synchronism with the clock of the clock generator for a certain duration of time in order to select a control signal **3**. As a result, a switching circuit **52** is turned on, thereby turning power on and causing the light-emitting diode **37** to emit light.

[0062] It is also possible to cause the two light-emitting diodes **36** and **37** to emit nearly simultaneously, rather than in a time-divided manner. In an example of a method of separately detecting light from a plurality of light sources, the individual light sources are driven by modulating them with different modulation frequencies. In this method, light from each light source can be separately detected by focusing on the frequency components contained in a detection signal from photodetectors.

[0063] The arrangement of the light-irradiating optical fibers, photodiodes, and the light-receiving optical fibers in the optical sensor portion **18** is determined based on the following theories (1) to (3).

[0064] (1) Regarding the position of the reflected-light receiving photodiode with respect to the light-irradiating optical fiber, it is most appropriate to position the light-receiving plane of the photodetector at a position where the reflected light is theoretically received, namely at a position within the plane of incidence of light on the subject where light reflected in a direction with an outgoing angle that is equal to the angle of incidence on a light-incident point of the subject is received. By locating the light-receiving plane of the reflected-light receiving photodiode at such a position, the ratio of reflected light in the amount of received light can be maximized.

[0065] (2) The scattered light-receiving photodetector has the light-receiving plane thereof disposed in a plane that forms an angle of approximately 90° with respect to the plane of incidence of light on the subject. The scattered-light receiving photodetector is disposed at approximately 90° relative to the reflected-light receiving photodetector because the source of light detected as scattered light is desired to be narrowed to the scattering phenomena as much as possible, as opposed to theory (1), or because the range of the phenomena as the object of detection of scattering is desired to be increased by the provision of the large angle of approximately 90°.

[0066] (3) The light-receiving end of the light-receiving optical fiber for traveled photon is disposed at a position in the plane of incidence of light on the subject that is farther than the light-receiving plane of the reflected-light receiving photodetector with respect to the light-irradiating optical fiber. The light-receiving end of the traveled-photon receiving optical fiber is thus disposed in the plane of incidence of light on the subject for the following reason. During the process in which light enters the skin and is scattered inside, the distribution of light spreads, and yet the distribution is greatest in the direction of incidence. As a result, the amount of light exiting from the skin is also greatest in this direction, so that the traveled photon can be most efficiently detected. Further, the light-receiving end of the traveled-photon receiving optical fiber is disposed farther than the light-receiving end of the reflected-light receiving photodetector with respect to the light-irradiating optical fiber. By so doing, a large amount of information can be detected, such as information relating to the absorption of light by hemoglobin in blood flowing in the capillary blood tubes during the process of light penetrating the skin and being scattered inside, or information relating to the thickness of skin, for example. It is also possible, however, to dispose the light-receiving optical fiber for traveled photon at a position other than that in the plane of incidence of light on the subject, though in that case the amount of traveled photon that is detected would be reduced.

[0067] In accordance with those theories (1) to (3), the exiting ends of the light-irradiating optical fibers, the photodetectors, and the receiving end of the light-receiving optical fiber are disposed in the optical sensor portion **18** as shown in the plan view of **FIG. 10A**. In this plan view, the light-irradiating optical fiber **31**, the reflected-light receiving photodiode **38** and the traveled-photon receiving optical fiber **33** are disposed substantially along an identical line XX. On a line YY with an angle α of approximately 90° with respect to the line XX connecting the light-irradiating optical fiber **31** and reflected-light receiving photodiode **38**, there are disposed the light-irradiating optical fiber **32** and the scattered-light receiving photodiode **39** for receiving scattered light from the light-irradiating optical fiber **31**. The light-irradiating optical fibers **31** and **32** and the photodiodes **38** and **39** are disposed more or less on an identical circle P about a center point at which the lines XX and YY intersect.

[0068] Regarding the angles of irradiation and detection of light by the light-irradiating optical fiber **31** and photodiode **38**, the reflected-light receiving photodiode **38** is positioned such that it can receive a beam of light reflected at a point (light incident point on the subject) y, namely the point y in **FIG. 8E**, that is located above the point of intersection of the lines XX and YY shown in **FIG. 10A** with the same angle as an incident angle θ that is formed by the axis of the light-irradiating optical fiber **31** and a normal to the surface of the subject at the incident point γ when the light emitted by the light-irradiating optical fiber **31** is reflected at the light incident point γ. Namely, the light-irradiating optical fiber **31** and the photodiode **38** are disposed such that the angles θ and ϕ are substantially identical.

[0069] By thus disposing the light-irradiating optical fiber **31**, the photodiode **38** and the traveled-photon receiving optical fiber **33** along the same line as shown in **FIG. 10A**, the amount of traveled photon detected by the light-receiving optical fiber **33** can be maximized. However, since the light-receiving optical fiber **33** is disposed in the same direction as the direction in which light is emitted from the light-irradiating optical fibers **31** and **32**, the ratio of reflected light or scattered light in the amount of received light increases. Further, as the photodiode **38**, the plate **21**, and the heat-conducting member **22** and thermistor **24** connected thereto are disposed along an identical line, the plate **21** and the heat-conducting member **22** and thermistor **24** connected thereto must be disposed away from the light-receiving optical fiber **33** along the line XX in order to allow the photodiode **38** to be disposed. As a result, the size of the optical sensor portion shown in **FIG. 10A** increases.

[0070] Alternatively, the exiting ends of the light-irradiating optical fibers, the photodiodes, and the receiving end of the light-receiving optical fiber may be disposed in the optical sensor portion **18** as shown in a plan view of **FIG. 10B**, in accordance with the theories (1) to (3). In the plan view, the light-irradiating optical fiber **31** and the reflected-light receiving photodiode **38** are disposed along the identical line XX. On a line YY with an angle of approximately 90° with respect to the line XX connecting the light-irradiating optical fiber **31** and reflected-light receiving photodiode **38**, there are disposed the light-irradiating optical fiber **32** and the light-receiving photodiode **39** for receiving scattered light from the light-irradiating optical fiber **31**. The traveled-photon receiving optical fiber **33** is disposed along a line ZZ that intersects the line YY at an angle α of approximately 45°. The light-irradiating optical fibers **31** and **32** and the photodiodes **38** and **39** are disposed more or less on an identical circle P about a center where the lines XX and YY intersect. Regarding the angles of irradiation and detection of light by the light-irradiating optical fiber **31** and the photodiode **38**, the reflected-light receiving photodiode **38** is positioned such that it can receive a beam of light reflected at a point (light incident point on the subject) γ, namely the point γ in **FIG. 8E**, that is located above the point of intersection of the lines XX and YY shown in **FIG. 10A** with the same angle as an incident angle θ that is formed by the axis of the light-irradiating optical fiber **31** and a normal to the surface of the subject at the incident point γ when the light emitted by the light-irradiating optical fiber **31** is reflected at the light incident point γ. Namely, the light-irradiating optical fiber **31** and the photodiode **38** are disposed such that the angles θ and +are substantially identical.

[0071] By thus disposing the traveled-photon receiving optical fiber **33** on the line ZZ as shown in **FIG. 10B**, though the amount of traveled photon that can be detected by the light-receiving optical fiber **33** decreases, the distance between the point of intersection of the lines XX and YY and the light-receiving optical fiber **33** can be reduced in the line ZZ direction. Accordingly, the size of the optical sensor portion **18** can be reduced. Further, as the light-receiving optical fiber **33** is disposed at a position approximately 45° away from the direction in which the light-irradiating optical fibers **31** and **32** radiate, the influence of reflected light or scattered light can be minimized, so that a large amount of traveled photon can be detected in the amount of received light.

[0072] Further alternatively, the exiting ends of the light-irradiating optical fibers, the photodiodes, and the receiving end of the light-receiving optical fiber in the optical sensor portion **18** may be disposed as shown in a plan view of **FIG. 10C**, in accordance with the theories (1) to (3). Namely, the light-irradiating optical fibers **31** and **32** and the photodiodes **38** and **39** may be disposed at any position on a circle P with a center β and a radius corresponding to the line between the center β and the light-irradiating optical fiber **31** as long as the relationship of the line XX intersecting the line YY at approximately 90° is maintained. For example, as shown in **FIG. 10C**, the optical sensor portion **18** may be configured in the following manner. The light-irradiating optical fiber **31** is disposed at a position corresponding to that of the photodiode **38** of **FIG. 10B**, and the photodiode **38** is disposed at a position corresponding to that of the light-irradiating optical fiber **31** of **FIG. 10B**. The light-irradiating optical fiber **32** is disposed at a position corresponding to that of the photodiode **39** of **FIG. 10B**, and the photodiode **39** is disposed at a position corresponding to that of the light-irradiating optical fiber **32** of **FIG. 10B**. The traveled-photon receiving optical fiber **33** is disposed on the line ZZ that intersects the line YY at approximately 45°. Regarding the angles of irradiation and detection of light by the light-irradiating optical fiber **31** and the photodiode **38**, the reflected-light receiving photodiode **38** is positioned such that it can receive a beam of light reflected at a point (light incident point on the subject) γ, namely the point γ in **FIG. 8E**, that is located above the point of intersection of the lines XX and YY shown in **FIG. 10C** with the same angle as an incident angle θ that is formed by the axis of the light-irradiating optical fiber **31** and a normal to the surface of the subject at the incident point γ when the light emitted by the light-irradiating optical fiber **31** is reflected at the light incident point γ. Namely, the light-irradiating optical fiber **31** and the photodiode **38** are disposed such that the angles θ and ϕ are substantially identical.

[0073] In this arrangement, the traveled-photon receiving optical fiber **33** is positioned in a direction opposite to that in which the light-irradiating optical fibers **31** and **32** radiate, so that, although the amount of light received by the light-receiving optical fiber **33** is fairly small, the received light contains hardly any reflected light or scattered light and consists mostly of traveled photon.

[0074] Regarding the arrangement of the light-irradiating optical fibers, photodiodes, and light-receiving optical fiber in the optical sensor portion **18** shown in **FIGS. 10A** to **10**C, a branch optical fiber **32***a* may be connected to the light-irradiating optical fiber **31**, and a light-emitting diode **37** may be disposed at the end of the optical fiber **32***a*, instead of using the light-irradiating optical fiber **32**. A top view of this arrangement of the optical fibers and the light-emitting diode is shown in **FIG. 10D**. **FIG. 10E** is a cross section taken along line XX of **FIG. 10D**, and **FIG. 10F** a cross section taken along line YY. The ZZ cross section of **FIG. 10D** is similar to that of **FIG. 8B**.

[0075] Regarding the optical sensor portion **18**, the exiting ends and the receiving planes of the light-irradiating optical fibers **31** and **32** and the photodiodes **38** and **39**, respectively, may be displaced a little in their optical axial directions as long as they are aimed at the light incident point γ on the subject (see **FIG. 8E**). In that case, the light-irradiating optical fibers **31** and **32** and the photodiodes **38** and **39** would not be all disposed on the identical circle P as shown, but would be displaced from one another in the vertical or height direction. However, if the light-irradiating optical fibers **31** and **32** are disposed at different heights, the intensity of irradiated light would be larger near the body surface and would be lower away from the body surface. Further, if the photodiodes **38** and **39** are disposed at different heights, the intensity of detected light would increase near the body surface and would decrease away from the body surface due to the spreading of light. Thus, such an arrangement would make it difficult to carry out measurement in a uniform environment and a correction of the information detected by the photodiodes would be necessary. In general, the light-irradiating optical fibers and the photodiodes are disposed near where light is irradiated so that an accurate measurement can be conducted. In the configuration of the present invention, the light-irradiating optical fibers and the photodiodes are disposed as close to the body surface as possible without hindering other functions for measurements, such as the measurement of temperatures. Further, the light-irradiating optical fibers **31** and **32** and the photodiodes **38** and **39** are disposed on the identical circle P or, more generally, near the plane of a cone whose apex is at the point y of incidence of light on the subject, such that a uniform environment for measuring radiated light and detected light can be obtained and an accurate measurement can be conducted.

[0076] Further regarding the optical sensor portion **18** shown in **FIGS. 10A** to **10**D, the traveled-photon receiving optical fiber **33** may be disposed at any point on a circle with a center β and a radius corresponding to the line connecting the center β and the light-receiving optical fiber **33**, as indicated by a dashed line in **FIG. 10G**. In this case, the distance between the exiting end (the light incident point) of the light-irradiating optical fiber and the receiving end of the traveled-photon receiving optical fiber **33** (where the light as the object of reception is incident) would be larger than the distance between the light incident point and the reception end of the photodiode **38** or the receiving end of the photodiode **39**. In such an arrangement, the placement of the traveled-photon receiving optical fiber **33** can be freely set, so that the optical sensor portion **18** can be configured in various ways as needed.

[0077] The photodiodes **38** and **39** provide reflectance R as measurement data, and absorbance can be approximately calculated from log(1/R). Light of wavelengths 810 nm and 950 nm is irradiated, and R is measured for each and log(1/R) is obtained for each, so that absorbance A_{D11} and A_{D21} at wavelength 810 nm and absorbance A_{D12} and A_{D22} at wavelength 950 nm can be measured. Part of the light penetrates into the skin and travels a certain distance d while being scattered therein repeatedly. The intensity I_{D3i} of traveled photon is measured by a photodiode **40**. (The absorbance of reflected light of wavelength λ_i detected by the photodiode for detecting reflected light is referenced by A_{D1i}, the absorbance of scattered light of wavelength λ_i detected by the photodiode for detecting scattered light is referenced by A_{D2i}, and the intensity of traveled photon of wavelength λ_i detected by the photodiode **40** is referenced by I_{D3i}.)

[0078] When the reduced hemoglobin concentration is [Hb] and the oxyhemoglobin concentration is [HbO_2], scattered-light absorbance A_{D2i} at wavelength λ_i is expressed by the following equations:

$$A_{D2i} = a\{[Hb] \times A_{Hb}(\lambda i) + [(HbO)_2] \times A_{HbO_2}(\lambda i)\} \times D \times a_R$$

$$a_R = \frac{b \times \sum A_{D2i}}{\sum A_{D1i}},\ D = \frac{1}{\frac{c \times \sum I_{D3i}}{i}}$$

where $A_{Hb}(\lambda_i)$ and $A_{Hb02}(\lambda_i)$ are the molar absorbance coefficients of the reduced hemoglobin and the oxyhemoglobin, respectively, and are known at the respective wavelengths. Terms a, b, and c are proportionality coefficients. A_{D1i} is the reflected-light absorbance at wavelength λ_i, and I_{D3i} is the traveled photon intensity at wavelength λ_i. From the above equations, the parameter a_R, which is determined by the relationship between reflected light and scattered light, and the parameter D of the skin thickness can be determined as constants, and can be substituted in the equation of A_{D2i}. The parameter determined by the relationship between reflected light and scattered light is a parameter relating to the roughness of the skin surface, for example, and the influence of the roughness of the skin surface, for example, can be corrected using that parameter. The parameter relating to the thickness of the skin can be determined from the measurement value obtained by the traveled-photon detector, and the influence of the thickness of the skin can be corrected using that parameter. Since i=2 wavelengths, two equations of A_{D2i} are produced. By solving these simultaneous equations, the two variables to be obtained, namely [Hb] and [HbO_2], can be obtained. The hemoglobin concentration [Hb]+[HbO_2], and the hemoglobin oxygen saturation [HbO_2]/([Hb]+[HbO_2]) can be determined from the above-obtained [Hb] and [HbO_2].

[0079] Although the present example has been described with regard to the measurement of the hemoglobin concentration and the hemoglobin oxygen saturation based on the measurement of absorbance at two wavelengths, absorbance may be measured by adding one or more wavelengths at which the difference in molar absorbance coefficient between the oxyhemoglobin and the deoxyhemoglobin is large so as to increase the measurement accuracy.

[0080] For example, when six wavelengths are used for measurement, any of the configurations shown in **FIGS. 10A** to **10C** may be employed for the arrangement of the light-irradiating optical fibers, photodiodes, and the light-receiving optical fiber in the optical sensor portion **18** in accordance with the theories (1) to (3). However, in the case of six wavelengths, while the ZZ cross-section of **FIG. 10B** corresponds to **FIG. 8B**, the XX cross-section of **FIG. 10B** corresponds to **FIG. 11A**, and the YY cross-section of **FIG. 10B** corresponds to **FIG. 11B**. To the light-irradiating optical fiber **31** are connected three branch optical fibers **31***a*, **31***b* and **31***c* each provided at the end thereof with light-emitting diodes **36***a*, **36***b* and **36***c*, respectively. Likewise, three branch optical fibers **32***a*, **32***b* and **32***c* are connected to the light-irradiating optical fiber **32**, and light-emitting diodes **37***a*, **37***b* and **37***c* are connected to the ends of the respective branch optical fibers. By thus connecting three branch optical fibers to one light-irradiating optical fiber in a bundled manner, the size of the optical sensor portion **18** can be reduced.

[0081] The light-emitting diode **36***a* emits light of 810 nm, light-emitting diode **36***b* light of 880 nm, light-emitting diode **36***c* light of 950 nm, light-emitting diode **37***a* light of 450 nm, light-emitting diode **37***b* light of 520 nm, and light-emitting diode **37***c* light of 660 nm, for example. Using the result of detection of irradiated light having these six wavelengths, corrections can be made for the influences of interfering components on the determination of hemoglobin concentration and hemoglobin oxygen saturation from absorbance, the interfering components including melanin pigment, bilirubin and the turbidity of blood, for example. Thus, the accuracy of measurement can be improved.

[0082] As mentioned above, by providing the light-irradiating optical fiber with three branches, an ideal configuration can be obtained in which the three light-emitting diodes share the same point of light irradiation. However, since the actual fiber-irradiated light has certain spread, the same function can be provided by employing a structure in which a light-irradiating fiber is provided to each light-emitting diode and the tips of the fibers are bundled. This structure, which can employ conventional fibers and can therefore be made inexpensively, is shown in **FIGS. 12A** to 12D. **FIG. 12A** shows a front view, **FIG. 12B** shows an XX cross-section, **FIG. 12C** shows a YY cross-section, and **FIG. 12D** shows a ZZ cross-section. As shown in **FIG. 12A**, a light-irradiating optical fiber **31** consists of three fibers bundled at the tips thereof. As shown in **FIG. 12B**, a photodiode **40** is disposed using a fiber **33** in order to allow for an efficient reception of traveled photon. As shown in **FIG. 12C**, optical fibers **31** are connected to light-irradiating light-emitting diodes **36**, and the tips of the fibers are bundled together, such that each optical fiber **31** emits light from a corresponding light-emitting diode. The photodiode **38** may be disposed at a further spaced-apart position if its housing has a lens function. **FIG. 12D** shows a similar configuration.

[0083] **FIG. 13** shows an example of connection and blocking of light between an light-emitting diode and an optical fiber. A light-emitting diode **36** is provided with an opening with a slightly larger diameter than the external diameter of an optical fiber **31**. The opening is filled with an adhesive **41** and then the optical fiber **31** is inserted and fixed therein such that the light emitted by the light-emitting diode **36** can be guided in the direction opposite to the face of the optical fiber. The opening also functions to position and fix the optical fiber **31** in place. It goes without saying that the intensity of irradiation increases as the depth of the opening is closer to the plane of the light-emitting diode element. The adhesive **41** should preferably be one that absorbs only a little amount of the irradiated light and have a refractive index that is close to that of the light-emitting diode and the optical fiber core. The light-emitting diode **37** and the traveled-photon receiving photodiode **40** have the same structures.

[0084] In practice, the light-emitting diodes **36** and **37** and the photodiode **40** are equipped with a light-blocking cap **42** for preventing the leakage of light to the outside and the reception of light from the outside. The light-blocking cap **42** is formed by a soft material, such as a silicon resin, so that it can be easily mounted on the light-emitting diode or optical fiber assembly. With regard to the photodiodes **38** and **39**, which are disposed inside the sensor portion, there might be no need to provide such a measure because they are disposed inside the sensor portion and are therefore already blocked against external light.

[0085] **FIG. 14** is a conceptual chart showing the flow of data processing in the apparatus using two wavelengths. The apparatus according to the present example is equipped with a thermistor **23**, a thermistor **24**, a pyroelectric detector **27**, a thermistor **28**, and three photodetectors formed by photodiodes **38** to **40**. The photodiodes **38** and **39** measure absorbance at wavelengths 810 nm and 950 nm. The photodiode **40** measures the intensity at wavelengths 810 nm and 950 nm. Thus, the apparatus is supplied with ten kinds of measurement values including temperature, thermal, and optical measurement data. In the case where the wavelength 880 nm is added for improving accuracy, the number of kinds of measurement values fed to the apparatus would be 13.

[0086] The seven kinds of analog signals are supplied via individual amplifiers A1 to A7 to analog/digital converters AD1 to AD7, where they are converted into digital signals. Based on the digitally converted values, parameters x_i (i=1, 2, 3, 4, 5) are calculated. The following are specific descriptions of x_i (where e_1 to e_5 are proportionality coefficients):

[0087] Parameter proportional to heat radiation

$$x_1=e_1\times(T_3)^4$$

[0088] Parameter proportional to heat convection

$$x_2=e_2\times(T_4-T_3)$$

[0089] Parameter proportional to hemoglobin concentration

$$x_3=e_3\times([Hb]+[Hb_2])$$

[0090] Parameter proportional to hemoglobin saturation

$$x_4 = e_4 \times \left(\frac{[HbO_2]}{[Hb] + [HbO_2]} \right)$$

[0091] Parameter proportional to blood flow volume

$$x_5 = e_5 \times \left(\frac{1}{t_{CONT} \times (S_1 - S_2)} \right)$$

[0092] Then, normalized parameters are calculated from mean values and standard deviations of parameter x_i obtained from actual data pertaining to large numbers of able-bodied people and diabetic patients. A normalized parameter X_i (where i=1, 2, 3, 4, 5) is calculated from each parameter x_i according to the following equation:

$$X_i = \frac{x_i - \bar{x}_i}{SD(x_i)}$$

where

[0093] x_i: parameter

[0094] $\bar{x}_i$: mean value of the parameter

[0095] $SD(x_i)$: standard deviation of the parameter

[0096] Using the above five normalized parameters, calculations are conducted for conversion into a glucose concentration to be eventually displayed. A program necessary for the processing calculations is stored in a ROM in the microprocessor built inside the apparatus. The memory area required for the processing calculations is secured in a RAM similarly built inside the apparatus. The results of calculation are displayed on the LCD.

[0097] The ROM stores, as a constituent element of the program necessary for the processing calculations, a function for determining glucose concentration C in particular. The function is defined as follows. C is expressed by the below-indicated equation (1), where a_i (i=0, 1, 2, 3, 4, 5) is determined from a plurality of pieces of measurement data in advance according to the following procedure:

(1) A multiple regression equation is created that indicates the relationship between the normalized parameters and the glucose concentration C.

(2) Normalized equations (simultaneous equations) relating to the normalized parameters are obtained from equations obtained by the least-squares method.

(3) Values of coefficient a_i (i=0, 1, 2, 3, 4, 5) are determined from the normalized equations and then substituted into the multiple regression equation.

[0098] Initially, the regression equation (1) indicating the relationship between the glucose concentration C and the normalized parameters X_1, X_2, X_3, X_4, and X_5 is formulated.

$$\begin{aligned} C &= f(X_1, X_2, X_3, X_4, X_5) \\ &= a_0 + a_1X_1 + a_2X_2 + a_3X_3 + a_4X_4 + a_5X_5 \end{aligned} \tag{1}$$

[0099] Then, the least-squares method is employed to obtain a multiple regression equation that would minimize the error with respect to a measured value C_i of glucose concentration according to an enzyme electrode method. When the sum of squares of the residual is E, E is expressed by the following equation (2):

$$\begin{aligned} E &= \sum_{i=1}^{n} d_i^2 \\ &= \sum_{i=1}^{n} (C_i - f(X_{i1}, X_{i2}, X_{i3}, X_{i4}, X_{i5}))^2 \\ &= \sum_{i=1}^{N} \{C_i - (a_0 + a_1X_{i1} + a_2X_{i2} + a_3X_{i3} + a_4X_{i4} + a_5X_{i5})\}^2 \end{aligned} \tag{2}$$

[0100] The sum E of squares of the residual becomes minimum when partial differentiation of equation (2) with respect to a_0, a**2**, . . . , a_5 gives zero. Thus, we have the following equations:

$$\begin{aligned} \frac{\partial E}{\partial a_0} &= -2\sum_{i=1}^{n} \{C_i - (a_0 + a_1X_{i1} + a_2X_{i2} + a_3X_{i3} + a_4X_{i4} + a_5X_{i5})\} = 0 \\ \frac{\partial E}{\partial a_1} &= -2\sum_{i=1}^{n} X_{i1}\{C_i - (a_0 + a_1X_{i1} + a_2X_{i2} + a_3X_{i3} + a_4X_{i4} + a_5X_{i5})\} = 0 \\ \frac{\partial E}{\partial a_2} &= -2\sum_{i=1}^{n} X_{i2}\{C_i - (a_0 + a_1X_{i1} + a_2X_{i2} + a_3X_{i3} + a_4X_{i4} + a_5X_{i5})\} = 0 \\ \frac{\partial E}{\partial a_3} &= -2\sum_{i=1}^{n} X_{i3}\{C_i - (a_0 + a_1X_{i1} + a_2X_{i2} + a_3X_{i3} + a_4X_{i4} + a_5X_{i5})\} = 0 \\ \frac{\partial E}{\partial a_4} &= -2\sum_{i=1}^{n} X_{i4}\{C_i - (a_0 + a_1X_{i1} + a_2X_{i2} + a_3X_{i3} + a_4X_{i4} + a_5X_{i5})\} = 0 \\ \frac{\partial E}{\partial a_5} &= -2\sum_{i=1}^{n} X_{i5}\{C_i - (a_0 + a_1X_{i1} + a_2X_{i2} + a_3X_{i3} + a_4X_{i4} + a_5X_{i5})\} = 0 \end{aligned} \tag{3}$$

[0101] When the mean values of C and X_1 to X_5 are C_{mean} and X_{1mean} to X_{5mean}, respectively, since $X_{imean}=0$ (i=1 to 5), equation (1) yields:

$$a_0 = C_{mean} - a_1X_{1mean} - a_2X_{2mean} - a_3X_{3mean} - a_4X_{4mean} - a_5X_{5mean} = C_{mean} \quad (4)$$

[0102] The variation and covariation between the normalized parameters are expressed by equation (5). Covariation between the normalized parameter X_i (i=1 to 5) and C is expressed by equation (6).

$$S_{ij} = \sum_{k=1}^{n} (X_{ki} - X_{imean})(X_{kj} - X_{jmean}) = \sum_{k=1}^{n} X_{ki}X_{kj} \quad (i, j = 1, 2, \ldots 5) \quad (5)$$

$$S_{iC} = \sum_{k=1}^{n} (X_{ki} - X_{imean})(C_k - C_{mean}) = \sum_{k=1}^{n} X_{ki}(C_k - C_{mean}) \quad (i = 1, 2, \ldots 5) \quad (6)$$

[0103] Substituting equations (4), (5), and (6) into equation (3) and rearranging yields a set of simultaneous equations (normalized equations) (7). Solving the set of equations (7) yields a_1 to a_5.

$$\begin{aligned} &a_1S_{11}+a_2S_{12}+a_3S_{13}+a_4S_{14}+a_5S_{15}=S_{1C} \\ &a_1S_{21}+a_2S_{22}+a_3S_{23}+a_4S_{24}+a_5S_{25}=S_{2C} \\ &a_1S_{31}+a_2S_{32}+a_3S_{33}+a_4S_{34}+a_5S_{35}=S_{3C} \\ &a_1S_{41}+a_2S_{42}+a_3S_{43}+a_4S_{44}+a_5S_{45}=S_{4C} \\ &a_1S_{51}+a_2S_{52}+a_3S_{53}+a_4S_{54}+a_5S_{55}=S_{5C} \quad (7) \end{aligned}$$

[0104] Constant term a_0 is obtained by means of equation (4). The thus obtained a_i (i=0, 1, 2, 3, 4, 5) is stored in ROM at the time of manufacture of the apparatus. In actual measurement using the apparatus, the normalized parameters X_1 to X_5 obtained from the measured values are substituted into regression equation (1) to calculate the glucose concentration C.

[0105] Hereafter, an example of the process of calculating parameter Xi will be described. The example concerns measurement values obtained from able-bodied persons. Coefficients for the parameter calculation equations are determined by temperature data and optical measurement data that have been measured in advance. The ROM in the microprocessor stores the following formula for the calculation of the parameter:

$$x_1 = 0.98 \times 10^{-3} \times (T_3)^4$$

$$x_2 = -1.24 \times (T_4 - T_3)$$

$$x_3 = 1.36 \times ([Hb] + [HbO_2])$$

$$x_4 = 2.67 \times \left(\frac{[HbO_2]}{[Hb] + [HbO_2]}\right)$$

$$x_5 = 1.52 \times 10^6 \times \left(\frac{1}{t_{CONT} \times (S_1 - S_2)}\right)$$

[0106] When T_3=36.5° C. is substituted in the above equations as a measurement value, for example, $x_1=1.74\times10^3$. When T_4=19.7° C. is substituted in the above equations, $x_2=2.08\times10$. Then, before finding X_3, it is necessary to find [Hb] and [HbO$_2$]. The coefficients for a concentration calculation formula are determined by the scattered-light absorbance coefficient of each substance that has been measured in advance. Using that formula, [Hb] and [HbO$_2$] can be determined by solving the following set of simultaneous equations in the case of measurement using two wavelengths:

$$\begin{aligned} A_{D2\_810} &= 1.86 \\ &= 0.87\{800 \times [Hb] + 1050 \times [HbO_2]\} \times 1.04 \times 0.85 \\ A_{D2\_950} &= 2.02 \\ &= 0.87\{750 \times [Hb] + 1150 \times [HbO_2]\} \times 1.04 \times 0.85 \\ a_R &= 0.85 \\ &= \frac{1.35 \times (1.67 + 1.98)}{(2.65 + 3.14)} \\ D &= 1.04 \\ &= \frac{1}{\frac{0.95 \times (1.02 + 1.01)}{2}} \end{aligned}$$

[0107] Solving this set of simultaneous equations gives [Hb]=0.17 mmol/L and [HbO$_2$]=2.17 mmol/L. Thus we have x_3=3.18 and $x_{4=2.48}$. Then, substituting $S_1=1.76\times10^2$, $S_2=1.89\times10$, and $t_{CONT}=^{22}$ seconds gives $x_5=4.40\times10^2$.

[0108] The hemoglobin concentration ([Hb]+[HbO$_2$]) was calculated to be 2.34 mmol/L. When the hemoglobin concentration was measured at the same time by an invasive method, i.e. by blood sampling, the value was 2.28 mmol/L.

[0109] When the traveled photon is not similarly detected by the light-receiving optical fiber **33** at the same time, the information about the parameter of the thickness of the skin would not be obtained. In that case, the below-indicated simultaneous equations would be obtained, and solving them would yield [Hb]=0.18 mmol/L and [HbO$_2$]=2.26 mmol/L. Thus, the hemoglobin concentration ([Hb]+[HbO$_2$]) would be 2.44 mmol/L.

$$\begin{aligned} A_{D2\_810} &= 1.86 \\ &= 0.87\{800 \times [Hb] + 1050 \times [HbO_2]\} \times 0.85 \\ A_{D2\_950} &= 2.02 \\ &= 0.87\{750 \times [Hb] + 1150 \times [HbO_2]\} \times 0.85 \\ a_R &= 0.85 \\ &= \frac{1.35 \times (1.67 + 1.98)}{(2.65 + 3.14)} \end{aligned}$$

[0110] Thus, it has been confirmed that the result of calculation in the case where traveled photon is detected by the light-receiving optical fiber **33** is closer to the value of hemoglobin concentration measured by blood sampling than the calculation result in the case where traveled photon is not detected by the light-receiving optical fiber **33**. Thus, it has been shown that the measurement accuracy can be improved by providing the optical sensor portion **18** with the light-receiving optical fiber **33**.

[0111] Next, X_1 to X_5 are obtained. X_1 to X_5 are the results of normalization of the above-obtained parameters X_1 to X_5. Assuming the distribution of a parameter is normal, 95% of a normalized parameter takes on values between −2 and +2. The normalized parameters can be determined by the following equations:

$$X_1 = -0.06 = \frac{1.74\times10^3 - 1.75\times10^3}{167}$$

$$X_2 = 0.04 = \frac{2.08\times10 - 2.06\times10}{5}$$

$$X_3 = 0.05 = \frac{3.18 - 3.15}{0.60}$$

$$X_4 = -0.12 = \frac{2.48 - 2.54}{0.50}$$

$$X_5 = 0.10 = \frac{4.40\times10^2 - 4.28\times10^2}{120}$$

[0112] From the above equations, we have normalized parameters $X_1=-0.06$, $X_2=+0.04$, X**3**=+0.05, $X_4=-0.12$, and $X_5=+0.10$.

[0113] Hereafter, an example of the process of calculating the glucose concentration will be described. The coefficients for regression equation (1) are determined in advance based on many items of data obtained from able-bodied persons and diabetics, and the ROM in the microprocessor stores the following formula for calculating the glucose concentration:

$$C=99.1+18.3\times X_1-20.2\times X_2-24.4\times X_3-21.8\times X_4-25.9\times X_5$$

[0114] Substituting X_1 to X_5 into the above equation gives C=96 mg/dl. Substituting normalized parameters $X_1=+1.15$, $X_2=-1.02$, $X_3=-0.83$, $X_4=-0.91$, and $X_5=-1.24$, which can be obtained as an example of the measured values for a diabetic patient, in the equation yields C=213 mg/dl.

[0115] The following describes the results of measurement by the conventional enzymatic electrode method in which a blood sample is reacted with a reagent and the amount of resultant electrons is measured to determine glucose concentration, and the results of measurement by an embodiment of the invention. When the glucose concentration for an able-bodied person was 89 mg/dl according to the enzymatic electrode method in one example, substituting the normalized parameters $X_1=-0.06$, $X_2=+0.04$, $X_3=+0.07$, $X_4=-0.10$, and $X_5=+0.10$, which were obtained by measurement at the same time according to the invention, into the above equation yields C=95 mg/dl. In another example, when the measured value of glucose concentration for a diabetic patient was 238 mg/dl according to the enzymatic electrode method, substituting the normalized parameters $X_1=+1.15$, $X_2=-1.02$, $X_3=-0.86$, $X_4=1.02$, and $X_5=-1.24$, which were obtained by measurement at the same time according to the invention, into the above equation yields C=216 mg/dl. The results thus indicated that the method according to the invention can provide highly accurate glucose concentration values.

[0116] **FIG. 15** shows a graph plotting glucose concentrations for a plurality of patients, with the vertical axis showing the calculated values of glucose concentration according to the invention, and the horizontal axis showing the measured values of glucose concentration according to the enzymatic electrode method. It is seen that a good correlation is obtained by measuring the oxygen supply volume and the blood flow volume according to the method of the invention (correlation coefficient=0.9394).

[0117] Thus, the invention makes it possible to determine blood sugar levels in a non-invasive measurement with similar levels of accuracy to the conventional invasive method.

[0118] All publication, patents, and patent applications cited herein are incorporated herein by reference to their entirety.

What is claimed is:

1. An optical measurement apparatus comprising:

a first light source for producing light of a first wavelength;

a first optical fiber for irradiating a light incident point on the surface of a subject with the light from said first light source;

a second light source for producing light of a second wavelength;

a second optical fiber for irradiating said light incident point on the surface of the subject with the light from said second light source in a direction different from that of the light of said first wavelength;

a first photodetector on which reflected light of the light of said first wavelength reflected by said light incident point and scattered light of the light of said second wavelength are incident without via fiber;

a second photodetector on which reflected light of the light of said second wavelength reflected at said light incident point and scattered light of the light of said first wavelength are incident without via fiber;

a third photodetector; and

a third optical fiber having an incident end thereof disposed at such a position as to be in contact with the surface of said subject, said third optical fiber being adapted to receive light exiting from an area spaced apart from said light incident point on the surface of said subject and then transmit the light to said third detector.

2. The optical measurement apparatus according to claim 1, wherein

said first and second light sources are adapted to emit light in a time-divided manner so that said light incident point on the surface of said subject is irradiated with the light of said first wavelength and the light of said second wavelength in a time-divided manner;

the reflected light of said first wavelength from said light incident point is mainly incident on said first photodetector when said first light source is emitting, while the scattered light of said second wavelength is mainly incident when said second light source is emitting; and

the reflected light of said second wavelength from said light incident point is mainly incident on said second photodetector when said second light source is emitting, and the scattered light of said first wavelength is mainly incident when said first light source is emitting.

3. The optical measurement apparatus according to claim 1, wherein a plane of incidence of the light of said first wavelength and a plane of incidence of the light of said second wavelength are substantially perpendicular to each other with respect to said light incident point on the surface of said subject.

4. The optical measurement apparatus according to claim 1, wherein an exiting end of said first optical fiber, an exiting end of said second optical fiber, a receiving plane of said first photodetector, and a receiving plane of said second photodetector are disposed in the vicinity of a circular conical surface with an apex thereof located at said light incident point on the surface of said subject.

5. The optical measurement apparatus according to claim 1, wherein the distance between said light incident point on the surface of said subject and the incident end of said third optical fiber is larger than the distance between the light incident point and the incident plane of said first photodetector and the receiving plane of said second photodetector.

6. The optical measurement apparatus according to claim 1, wherein the incident end of said third optical fiber is located on the plane of incidence of the light of said first wavelength or on the plane of incidence of the light of said second wavelength.

7. The optical measurement apparatus according to claim 1, wherein said first wavelength is a wavelength at which the molar absorption coefficient of oxyhemoglobin and that of reduced hemoglobin are equal, and said second wavelength is a wavelength used for the detection of a difference in absorbance between oxyhemoglobin and reduced hemoglobin.

8. The optical measurement apparatus according to claim 1, wherein a measurement error due to the thickness of skin is corrected using optical intensity measured by said third detector.

9. The optical measurement apparatus according to claim 1, wherein a branch optical fiber is connected to said first optical fiber and/or said second optical fiber, and a light source for emitting light of a wavelength different from that of said first light source and said second light source is provided at an end of said branch optical fiber.

10. The optical measurement apparatus according to claim 1, wherein reflected light of said first wavelength reflected at said light incident point and scattered light of said second wavelength are directly incident on said first photodetector, and reflected light of said second wavelength reflected at said light incident point and scattered light of said first wavelength are directly incident on said second photodetector.

11. A blood sugar level measuring apparatus comprising:

(1) a heat-amount measuring portion for measuring a plurality of temperatures deriving from a body surface and acquiring information that is used for calculating a convective heat transfer amount and radiation heat transfer amount that are related to the dissipation of heat from the body surface;

(2) a blood flow volume measuring portion for acquiring information about the volume of blood flow;

(3) an optical measurement portion including a light source for producing light of at least two different wavelengths, an optical system for irradiating the body surface with the light emitted by said light source, and at least three different photodetectors for detecting the light that has been irradiated onto the body surface, said optical measurement portion providing hemoglobin concentration and hemoglobin oxygen saturation in blood;

(4) a memory portion in which relationships between parameters respectively corresponding to said plurality of temperatures, blood flow volume, hemoglobin concentration and hemoglobin oxygen saturation in blood, and blood sugar levels are stored;

(5) a calculation portion for converting a plurality of measurement values inputted from said heat amount measuring portion, said blood flow volume measuring portion, and said optical measurement portion respectively into said parameters, and then calculating a blood sugar value by applying said parameters to said relationships stored in said memory portion; and

(6) a display portion for displaying the blood sugar level calculated by said calculation portion, wherein

said optical measurement portion includes a first light source for producing light of a first wavelength, a second light source for producing light of a second wavelength, a first optical fiber, a second optical fiber, a third optical fiber, a first photodetector, a second photodetector, and a third photodetector,

a light incident point on the surface of a subject is irradiated with light emitted by said first light source via said first optical fiber,

said light incident point on the surface of said subject is irradiated with light emitted by said second light source via said second optical fiber in a direction different from that of the light of said first wavelength, reflected light of the light of said first wavelength reflected at said light incident point and scattered light of the light of said second wavelength are incident on said first photodetector without via fiber,

reflected light of the light of said second wavelength reflected at said light incident point and scattered light of the light of said first wavelength are incident on said second photodetector without via fiber,

said third optical fiber has an incident end thereof disposed at such a position as to be in contact with the surface of said subject in an area spaced apart from said light incident point on the subject surface, and

said third photodetector is adapted to receive light exiting from an area spaced apart from said light incident point on the surface of the subject via said third optical fiber.

12. The blood sugar level measuring apparatus according to claim 11, wherein a plane of incidence of the light of said first wavelength and a plane of incidence of the light of said second wavelength are substantially perpendicular to each other with respect to said light incident point on the surface of said subject.

13. The blood sugar level measuring apparatus according to claim 11, wherein an exiting end of said first optical fiber, an exiting end of said second optical fiber, a receiving plane of said first photodetector, and a receiving plane of said second photodetector are disposed in the vicinity of a circular conical surface with an apex located at said light incident point.

14. The blood sugar level measuring apparatus according to claim 11, wherein an incident end of said third optical fiber is located on a plane of incidence of the light of said first wavelength, or a plane of incidence of the light of said second wavelength.

15. The blood sugar level measuring apparatus according to claim 11, wherein an incident end of said third optical fiber is located on a plane that forms an angle of approximately 45° with a plane of incidence of the light of said first wavelength and with a plane of incidence of the light of said second wavelength.

16. The blood sugar level measuring apparatus according to claim 11, wherein said first wavelength is a wavelength at which the molar absorption coefficient of oxyhemoglobin and that of reduced hemoglobin are equal, and said second wavelength is a wavelength for detecting a difference in absorbance between oxyhemoglobin and reduced hemoglobin.

17. The blood sugar level measuring apparatus according to claim 11, wherein

said optical measurement portion further comprises a control portion for controlling the emission of light from said first light source and said second light source, said control portion causing said first light source and said second light source to emit light alternately such that said light incident point on the surface of the subject is irradiated with the light of said first wavelength and the light of said second wavelength alternately,

reflected light of said first wavelength from said light incident point is mainly incident on said first photodetector when said first light source is emitting, and scattered light of the light of said second wavelength is mainly incident when said second light source is emitting,

reflected light of said second wavelength from said light incident point on the surface of the subject is mainly incident on said second photodetector when said second light source is emitting, and scattered light of the light of said first wavelength is mainly incident when said first light source is emitting.

18. The blood sugar level measuring apparatus according to claim 11, wherein the reflected light of the light of said first wavelength reflected at said light incident point and the scattered light of the light of said second wavelength are directly incident on said first photodetector, and the reflected light of the light of said second wavelength reflected at said light incident point and the scattered light of the light of said first wavelength are directly incident on said second photodetector.

19. A blood sugar level measuring apparatus comprising:

an ambient temperature detector for measuring ambient temperature;

a body-surface contact portion with which a body surface comes into contact;

a radiation temperature detector for measuring radiation heat from said body surface;

an adjacent temperature detector disposed adjacent to said body-surface contact portion;

an indirect temperature detector for detecting the temperature at a position spaced apart from said body-surface contact portion;

a heat conducting member connecting said body-surface contact portion and said indirect temperature detector;

a light source for producing light of at least two different wavelengths;

an optical system for irradiating the body surface with light emitted by said light source;

at least three different photodetectors for detecting the light that has been irradiated onto the body surface;

a memory portion in which relationships between individual outputs from said ambient temperature detector, said radiation temperature detector, said adjacent temperature detector, said indirect temperature detector, and said at least three different photodetectors, and blood sugar levels are stored;

a calculation portion for calculating a blood sugar level by applying said individual outputs to said relationships stored in said memory portion; and

a display portion for displaying the result of calculation in said calculation portion, wherein

said light source comprises a first light source for producing light of a first wavelength and a second light source for producing light of a second wavelength,

said optical system comprises a first optical fiber and a second optical fiber,

said photodetectors comprises a first photodetector, a second photodetector, and a third photodetector that receives light via a third optical fiber, wherein

light emitted by said first light source is irradiated onto a light incident point on the surface of a subject via said first optical fiber,

light emitted by said second light source is irradiated onto said light incident point on the subject surface via said second optical fiber in a direction different from the light of said first wavelength,

reflected light of the light of said first wavelength reflected at said light incident point and scattered light of the light of said second light are incident on said first photodetector without via fiber,

reflected light of the light of said second wavelength reflected at said light incident point and scattered light of the light of said first wavelength are incident on said second photodetector without via fiber,

said third optical fiber has an incident end thereof located at such a position as to be in contact with the subject surface in an area spaced apart from said light incident point on the subject surface, and said third photodetector is adapted to receive, via said third optical fiber, light exiting from an area spaced apart from said light incident point on the subject surface.

20. The blood sugar level measuring apparatus according to claim 19, wherein the reflected light of the light of said first wavelength reflected at said light incident point and the scattered light of the light of said second wavelength are directly incident on said first photodetector, and the reflected light of the light of said second wavelength reflected at said light incident point and the scattered light of the light of said first wavelength are directly incident on said second photodetector.

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专利名称(译)	光学测量装置和使用其的血糖水平测量装置		
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摘要(译)

一种用于基于温度测量非侵入性地测量血糖水平的装置。通过温度测量方法非侵入性地测量的血糖水平通过血氧饱和度和血流量来校正。光学传感器在穿透皮肤后检测散射光，反射光和从体表射出的光，从而可以通过考虑皮肤厚度对血氧饱和度的影响来稳定测量数据。