A METHOD OF OBTAINING A CORRECTION FACTOR AND A CALIBRATION DEVICE FACILITATING PERFORMANCE OF SAID METHOD

TECHNICAL FIELD

The present invention relates to a method of obtaining a correction factor to compensate for wavelength shifts of a light source and a calibration device facilitating the performance of said method.

BACKGROUND

Photo-plethysmography (PPG) deals with measuring the modulation of optical signals transmitted through the human body and relating these signals to a particular physiological condition. A use of PPG is in pulse oximetry which utilizes PPG signals to evaluate the concentration of oxygenated hemoglobin with respect to total hemoglobin in a person's blood stream. One of the popular devices utilizing this concept is the finger pulse oximeter.

In a pulse oximeter, there are two main components: a light emitting diode (LED) and a photo-detector (PD). Light from the LED passes through a person's skin and interacts with the blood flowing in the person's artery. As the blood flow changes mainly in the artery of a person's finger due to the pulsations of the heart, the signal received by the photo-detector modulates in accordance with the varying blood-flow in the artery. Generally, there are two LEDs used, with varying absorption coefficients of oxygenated (HbO_2) and de-oxygenated hemoglobin (Hb), to determine the ratiometric concentration of HbO_2 with respect to total hemoglobin. This ratio is determined as a percentage called SpO_2 value which provides a measurement of the oxygen saturation of a person's blood.

There are, in general, two forms of finger pulse-oximeters: Transmission type and Reflection type. In the transmission based pulse-oximeter, light from the LEDs passes through the skin, capillaries, bones, etc and reaches the detector which measures the transmittance. However, in the reflectance based pulse-oximeter, the LEDs and the photo-detector are in substantially the same plane, and light from the LEDs reach the photo-detector through scattering and reflection paths. In both these cases, the signals received by the photo-detector and the accordingly derived SpO_2 value depends strongly on the wavelength of the LED as further explained below.

The light path from the LEDs to the PD is significantly complicated due to absorption and scattering at various layers of the skin, fat, capillaries, etc. There is generally an empirical relationship between the signals received by the PD and the SpO_2 value. This empirical relationship is determined by tests conducted on various test subjects with varying SpO_2 values. The empirical relationship can therefore be defined by a saturation curve.

The saturation curve of the pulse oximeter is dependent on the respective wavelengths of the source LEDs (i.e. wavelength specific). Thus, if a wrong saturation curve is relied upon after there has been a wavelength shift of the source LED, an error in the determination of the SpO_2 value can result. In practice, the wavelength of a LED varies over time due to many factors attributed towards ageing such as residual mechanical stress relaxation, material diffusion, etc resulting in the inaccuracies in the measured SpO_2 values when only a single saturation curve is relied upon.

However, typical oximetry devices do not take into account the possibility of changes in the LED's wavelength over time and/or do not utilize more than one saturation curve when

measuring SpO_2 values, resulting in less than satisfactory accuracy in the devices, especially after a certain period of use.

Therefore, the inventors have recognized that, ideally, there should be different saturation curves for different wavelengths of LEDs. Further, if there is a change in the wavelength of the LEDs over time, an appropriate saturation curve can be chosen to determine a more accurate SpO_2 value.

In view of the above, there is a need for a method of obtaining a correction factor to compensate for wavelength shifts of the light source of oximetry devices. In particular, there is a need to measure the change in the wavelength of the source LED from its value at the time of manufacturing so that an appropriate saturation curve can be utilized for accurate determination of the subject's SpO_2 value. In addition, there is also a need for a calibration device capable of facilitating the performance of the method of obtaining a correction factor to compensate for wavelength shifts of the light source oximetry devices.

SUMMARY

According to one aspect, there is provided a method of obtaining a correction factor to compensate for wavelength shifts of a light source, the method comprising (i) detecting by a light detector, a first optical characteristic of substantially unfiltered light emitted by the light source; (ii) detecting by the light detector, a second optical characteristic of filtered light from the light source; and (iii) comparing the first optical characteristic, the second optical characteristic and a predetermined set of values to obtain the correction factor. The light source and the light detector may be comprised in a physiological parameter measurement device.

In one embodiment, the comparing step (iii) comprises obtaining a ratio of the first optical characteristic to the second optical characteristic. The comparing step (iii) may further comprise comparing the ratio with a predetermined ratio to obtain a value of the wavelength shift of the light source; and corresponding the value of the wavelength shift with a table of predetermined values to obtain the correction factor. The predetermined values of the table of predetermined values may be provided at incremental wavelengths. The predetermined ratio may be a ratio of the first optical characteristic to the second optical characteristic obtained at the time of manufacturing the light source by performing steps (i) to (ii). The predetermined ratio can be a ratio of the first optical characteristic to the second optical characteristic obtained at the time of manufacturing the physiological parameter measurement device by performing steps (i) to (ii). In one embodiment, the first and second optical characteristics are values of the power of the light detected by the light detector.

In one embodiment, the filtered light is produced by passing light from the light source through a filter. The filter may be an optical edge filter. The physiological parameter measurement device may comprise a photoplethysmography (PPG) sensing device. The photoplethysmography (PPG) sensing device may comprise a pulse oximeter. The light detector may comprise a photodetector. The light can be selected from the group consisting of X-rays, visible light rays, ultraviolet light rays and infra-red rays.

In one embodiment, wherein the ratio is represented by the formula

$$\frac{P_{\det}}{P_0} = \frac{f_2}{f_1} \frac{\int_{\lambda} T(\lambda) \cdot P_{LED}(\lambda) d\lambda}{\int_{\lambda} P_{LED}(\lambda) d\lambda},$$

wherein P_0 represents the power of the filtered light detected by the light detector, P_{det} represents the power of the substantially unfiltered light detected by the light detector, $T(\lambda)$ represents the transmission spectrum of the light source as a function of the wavelength λ, $P_{LED}(\lambda)$ represents the radiation spectrum of the light source as a function of the wavelength

λ, and f_1 and f_2 represent constants which are based on the light path of the substantially unfiltered and filtered light respectively.

In one embodiment, the predetermined ratio is represented by the formula

$$\frac{P_{det0}}{P_{i0}} = \frac{f_2}{f_1} \frac{\int_{\lambda} T(\lambda) \cdot P_{LED}(\lambda) d\lambda}{\int_{\lambda} P_{LED}(\lambda) d\lambda},$$

wherein P_{i0} represents the power of the filtered light detected by the light detector at the time of manufacturing the light source or the physiological parameter measurement device, P_{det0} represents the power of the substantially unfiltered light detected by the light detector at the time of manufacturing the light source or the physiological parameter measurement device, T(λ) represents the transmission spectrum of the light source as a function of the wavelength λ, $P_{LED}(\lambda)$ represents the radiation spectrum of the light source as a function of the wavelength λ, and f_1 and f_2 represent constants which are based on the light path of the substantially unfiltered and filtered light respectively.

According to another aspect, there is provided a calibration device for coupling to a physiological measurement device having a light source and a light detector, the calibration device comprising a wave guiding structure configured to allow light to travel from the light source to the light detector along a predetermined path, when the calibration device is coupled to said physiological measurement device; and a filter for producing filtered light from the light source when the calibration device is coupled to said physiological measurement device, the filter configured to be disposed in or out of the path of light travelling from the light source to the light detector, wherein the calibration device allows the physiological measurement device to perform the method disclosed herein when coupled thereto.

In one embodiment, the wave guiding structure comprises at least two separate paths, each of said separate path allowing light to travel from the light source to the light detector when in optical communication with the light source. At least two separate paths may be configured to be in interchangeable optical communication with the light source. The wave guiding structure may comprise a plurality of mirrors. A reflective plane of at least one mirror may be positioned at angle of about 45° to an emission plane of the light source and a reflective plane of at least another mirror is positioned at an angle of about 45° to an incident plane of the light detector, the emission plane being substantially perpendicular to the direction of the emitted light from the light source and the incident plane being substantially perpendicular to the direction of light incident on the light detector. The at least two separate paths may be intersect with each other at an angle of about 90°.The calibration device may further comprise a light shielding cover for shielding ambient light from the light detector when the calibration device is coupled to the physiological measurement device. In one embodiment, the filter is disposed on a rotatable disc. The wave guiding structure may be disposed on a rotatable disc. The filter may be an optical edge filter. The physiological parameter measurement device may comprise a photoplethysmography (PPG) sensing device. In one embodiment, the photoplethysmography (PPG) sensing device comprises a pulse oximeter. The light detector may comprise a photodetector. The light can be selected from the group consisting of X-rays, visible light rays, ultraviolet light rays and infra-red rays.

According to another aspect, there is provided a physiological parameter measurement device for performing the method disclosed herein, the measurement device comprising a light source for emitting light; a light detector for measuring a optical characteristic of light emitted by the light source; and a filter for producing filtered light from the light source, the filter configured to be interchangeably disposed in or out of the path of light travelling from the light source to the light detector.

It will be appreciated that the representative alphabets/symbols "P_0", "P_{det}", "$\Delta\lambda$", "$\Delta\lambda_{th}$", "K_0", "K" etc or the like are non-limiting alphabets/symbols that may be used to represent a particular parameter merely for easier illustrations. For example, the method of obtaining a correction factor to compensate for wavelength shifts of a light source, may comprise (i) detecting by a light detector, an optical characteristic P_0 of substantially unfiltered light emitted by the light source; (ii) detecting by the light detector, an optical characteristic P_{det} of filtered light from the light source; and (ii) comparing P_0, P_{det} and a predetermined set of values to obtain the correction factor. The comparing step (iii) may comprise obtaining a ratio K of the optical characteristic P_{det} to the optical characteristic P_0. The comparing step (iii) may further comprises comparing the ratio K with a predetermined value K_0 to obtain a value of the wavelength shift $\Delta\lambda$ of the light source; and corresponding the value of the wavelength shift $\Delta\lambda$ with a table of predetermined values to obtain the correction factor. The predetermined values of the table of predetermined values may be provided at incremental wavelengths $\Delta\lambda_{th}$. The predetermined value K_0 can be a ratio of P_{det} to P_0 obtained at the time of manufacturing the light source by performing steps (i) to (ii). The predetermined value K_0 may be a ratio of P_{det} to P_0 obtained at the time of manufacturing the physiological parameter measurement device by performing steps (i) to (ii). The optical characteristics P_0 and P_{det} may be values of the power of the light detected by the light detector. In some cases, other suitable alphabets may also be used for representation of similar parameters as and when appropriate.

BRIEF DESCRIPTION OF THE DRAWINGS

Example embodiments of the invention will be better understood and readily apparent to one of ordinary skill in the art from the following written description, by way of example only, and in conjunction with the drawings, in which:

Fig. 1 depicts graphs showing the variation in the SpO_2 vs. R for different wavelengths around the RED LED.

Fig. 2 depicts graphs showing variation in calculated SpO_2 values as a function of LED wavelength for different R values.

Fig. 3 is a transmission spectrum of an exemplary optical edge filter disclosed herein.

Fig. 4 is a power spectrum of a LED after transmission through an optical edge filter in accordance with an exemplary embodiment disclosed herein.

Fig. 5 is a graph of empirical results showing variations in the ratio of the detected power from an LED in the absence and presence of the optical edge filter and its variation with peak wavelength of the LED.

Fig. 6 is an isometric view of the different parts of a wavelength calibration device in accordance with an exemplary embodiment disclosed herein.

Fig. 7a is a schematic showing light trajectory from a LED to a photo-detector without the optical edge filter in between in accordance with the exemplary embodiment shown in Fig. 6.

Fig. 7b is a schematic showing light trajectory from the LED of Fig. 7a to the photo-detector of Fig. 7a with the optical edge filter in between in accordance with an exemplary embodiment disclosed herein.

Fig. 8 shows a typical wavelength spectrum of an LED approximated to a Lorentzian function.

Fig. 9 is a flowchart showing various steps carried out and data used during manufacturing in order to implement a calibration routine during regular use of a pulse oximeter in accordance with an exemplary embodiment disclosed herein.

Fig. 10 is a general flow-chart showing the steps taken to execute a calibration process in accordance with an exemplary embodiment disclosed herein.

Fig. 11 is an isometric view of the different parts of a wavelength calibration device in accordance with an exemplary embodiment disclosed herein.

Fig. 12 is a flowchart showing the utilization of different SpO_2 Vs. R curves for different wavelengths for compensation of wavelength shift in a calibration device in accordance with an exemplary embodiment disclosed herein.

Fig. 13 is a flowchart showing the utilization of a look-up table for wavelength variation of extinction coefficients for ε_{Hb} and ε_{HbO2} values as a compensation mechanism for a calibration device in accordance with an exemplary embodiment disclosed herein.

Fig. 14a is a flowchart showing the utilization of two mutually perpendicular waveguiding paths of a wavelength calibration device in accordance with an exemplary embodiment disclosed herein.

Fig. 14b is a flowchart showing the utilization of a single waveguiding path and a rotatable optical filter disk of a calibration device in accordance with an exemplary embodiment disclosed herein.

Fig 15a is an isometric view of mutually perpendicular waveguiding structures in accordance with an exemplary embodiment disclosed herein.

Fig. 15b is a top view of mutually perpendicular waveguiding structures of Fig. 15a.

DETAILED DESCRIPTION OF DRAWINGS

Non-limiting example embodiments of a method of obtaining a correction factor to compensate for wavelength shifts of a light source and a calibration device for facilitating the performance of said method are disclosed in the following paragraphs.

The description also discloses relevant device/apparatus for performing the steps of the described methods. Such apparatus may be specifically constructed for the purposes of the methods, or may comprise a general purpose computer/processor or other device selectively activated or reconfigured by a computer program stored in a storage member. The algorithms and displays described herein are not inherently related to any particular computer or other apparatus. It is understood that general purpose devices/machines may be used in accordance with the teachings herein. Alternatively, the construction of a specialized device/apparatus to perform the method steps may be desired.

In addition, it is submitted that the description also implicitly covers a computer program, in that it would be clear that the steps of the methods described herein may be put into effect by computer code. It will be appreciated that a large variety of programming languages and coding can be used to implement the teachings of the description herein. Moreover, the computer program if applicable is not limited to any particular control flow and can use different control flows without departing from the scope of the invention.

Furthermore, one or more of the steps of the computer program if applicable may be performed in parallel and/or sequentially. Such a computer program if applicable may be stored on any computer readable medium. The computer readable medium may include storage devices such as magnetic or optical disks, memory chips, or other storage devices suitable for interfacing with a suitable reader/general purpose computer. The computer readable medium may even include a wired medium such as exemplified in the Internet system, or wireless medium such as exemplified in bluetooth technology. The computer program when loaded and executed on a suitable reader effectively results in an apparatus that can implement the steps of the described methods.

The example embodiments may also be implemented as hardware modules. A module is a functional hardware unit designed for use with other components or modules. For example, a module may be implemented using digital or discrete electronic components, or it can form a portion of an entire electronic circuit such as an Application Specific Integrated Circuit (ASIC). A person skilled in the art will understand that the example embodiments can also be implemented as a combination of hardware and software modules.

In certain preferred example embodiments, a calibration device is provided which utilizes a single light source such as a photodiode, a single light detector such as a photodetector, one or more waveguiding structures and a filter in the form of an optical edge filter for calibrating the SpO_2 curve of a reflective pulse oximetry device. Advantageously, as only a single light source and a single light detector is used for calibration, the calibration procedure can be carried out in a relatively simple, straightforward and compact manner.

In some example embodiments, there is provided a method of obtaining a correction factor to compensate for wavelength shifts of a light source, the method comprising (i) detecting by a light detector, an optical characteristic P_0 of substantially unfiltered light

emitted by the light source; (ii) detecting by the light detector, an optical characteristic P_{det} of the filtered light from the light source; and (iii) comparing P_0, P_{det} and a predetermined set of values to obtain the correction factor.

In other example embodiments, there is provided a calibration device for coupling to a physiological measurement device having a light source and a light detector, the calibration device comprising a wave guiding structure configured to allow light to travel from the light source to the light detector along a predetermined path, when the calibration device is coupled to said physiological measurement device; and a filter for producing filtered light from the light source when the calibration device is coupled to said physiological measurement device, the filter configured to be disposed in or out of the path of light travelling from the light source to the light detector, wherein the calibration device allows the physiological measurement device to perform the method disclosed herein when coupled thereto.

In certain embodiments, the calibration features may be incorporated into a physiological parameter measurement device. Accordingly, in some example embodiments, there is provided a physiological parameter measurement device for performing the method disclosed herein, the measurement device comprising a light source for emitting light; a light detector for measuring a optical characteristic P_0 of light emitted by the light source; and a filter for producing filtered light from the light source, the filter configured to be interchangeably disposed in or out of the path of light travelling from the light source to the light detector.

It will be appreciated that the light source described herein is not restricted to a LED although in some exemplary embodiments, LED may be the preferred light source. Other suitable light sources may also be used. Likewise, the light detector described herein is not restricted to a photodetector, although in some exemplary embodiments, a photodetector

may be the preferred light detector. Other suitable light detectors may also be used. The filter described herein is also not meant to be limited strictly to an optical edge filter. Other suitable filters may also be used as appropriate.

It should also be noted that the physiological parameter measurement device described herein is not limited to a specific or particular kind of physiological parameter measurement device although in certain embodiments, a photoplethysmography (PPG) sensing device such as an oximeter is preferred.

The term "light" as used herein is meant to be interpreted in a broad sense and is not limited to visible light only. The term "light" as used herein can include but is not limited to X-rays, visible light rays, ultraviolet light rays and infra-red rays.

The following description of embodiments aims to provide a better understanding on the derivation and general principles of the method of obtaining a correction factor and a calibration device capable of the same disclosed herein. The following description of embodiments may from time to time draw references to the Figures for illustrative purposes. Accordingly, it is reiterated that the following description in conjunction with any of its referenced Figures should not be construed as being limiting or restrictive.

In an exemplary embodiment, the signal obtained for a SpO_2 measurement is a ratio of measurements taken at two different wavelengths. In this exemplary embodiment, the two wavelengths chosen are generally in the Red and Infra-Red regions of the electromagnetic spectrum. The ratio of the signals from the Red and Infra-Red LED's can be calculated as:

$$R = \frac{\ln\left(\frac{I_{max}}{I_{min}}\right)\Big|_{Red}}{\ln\left(\frac{I_{max}}{I_{min}}\right)\Big|_{IR}} \qquad (1)$$

Where, I_{MAX} and I_{MIN} are the AC and DC amplitudes of the Red (or RED) and Infra-Red (IR) signals respectively. From a more fundamental viewpoint, the R value is a function of the extinction coefficient of oxygenated and de-oxygenated blood at both the RED and IR wavelengths of the source LEDs. A detailed expression for R can be written as:

$$R = \frac{-\Delta c \cdot d_R \cdot [S \cdot \varepsilon_{HbO2}(\lambda_R) + (1-S) \cdot \varepsilon_{Hb}(\lambda_R)]}{-\Delta c \cdot d_{IR} \cdot [S \cdot \varepsilon_{HbO2}(\lambda_{IR}) + (1-S) \cdot \varepsilon_{Hb}(\lambda_{IR})]} \quad (2)$$

In the above equation Δc is the change in the concentration of hemoglobin during the systolic and diastolic stroke, ε_{HbO2} and ε_{Hb} are the extinction coefficients of oxygenated and deoxygenated hemoglobin respectively and d_R and d_{IR} are the path lengths of the light inside the skin from the source LED to the detector. Also, S in the above expression represents the SpO_2 value.

By rearranging the above expression, the SpO_2 value can be expressed as a function of R as follows:

$$SpO_2(\lambda_R, \lambda_{IR}) = \frac{\varepsilon_{Hb}(\lambda_R) \cdot d_R - R \cdot d_{IR} \cdot \varepsilon_{Hb}(\lambda_{IR})}{d_R \cdot \varepsilon_{Hb}(\lambda_R) - d_R \cdot \varepsilon_{HbO2}(\lambda_R) + R \cdot d_{IR} \cdot [\varepsilon_{HbO2}(\lambda_{IR}) - \varepsilon_{Hb}(\lambda_{IR})]} \quad (3)$$

The above expression is an analytically derived expression for the SpO_2 value as a function of source LED wavelengths. It is observed from the above expression that calculation of the SpO_2 value depends on the source wavelengths through the wavelength dependency of the extinction coefficients of oxygenated and deoxygenated hemoglobin. The variation in the SpO_2 as a function of R is shown for different wavelengths of the RED source LED, λ_R in Fig. 1 by utilizing equation (3) with known values for $\varepsilon_{Hb}(\lambda_R)$ and $\varepsilon_{HbO2}(\lambda_R)$.

Fig. 2 depicts a plot showing the variation of SpO_2 as a function of source LED wavelength variation for different R values in an exemplary embodiment. It can be observed from Fig. 2 that the variation in the calculation of SpO_2 value is more severe for larger R values than smaller R values.

Thus, based on the above and without being bound by theory, it is believed that the inclusion of a wavelength calibration device can improve the accuracy of SpO_2 measurement over an extended range, for example from 80 – 100%.

A simplified form of the above expression (3) is utilized in some commercial pulse oximeters whose saturation equation can be written as:

$$SpO_2 = R \cdot A + B \qquad (4)$$

Where, A and B are empirically determined constants. This expression is a straight line approximation to the previous SpO_2 expression in equation (3).

Assuming that due to a change of the wavelength of the RED LED, the emitted wavelength is λ_{R2} where

$$\lambda_{R2} = \lambda_R + \Delta\lambda_R \qquad (5)$$

The above expression (3) can then be written as:

$$SpO_2(\lambda_{R2}) = \frac{\varepsilon_{Hb}(\lambda_{R2}) \cdot d_R - R(\lambda_{R2}) \cdot d_{IR} \cdot \varepsilon_{Hb}(\lambda_{IR})}{d_R \cdot \varepsilon_{Hb}(\lambda_{R2}) - d_R \cdot \varepsilon_{HbO2}(\lambda_{R2}) + R(\lambda_{R2}, \lambda_{IR}) \cdot d_{IR} \cdot [\varepsilon_{HbO2}(\lambda_{IR}) - \varepsilon_{Hb}(\lambda_{IR})]} \qquad (6)$$

In the above expression, it is assumed that the change in optical path length, d_R, due to the wavelength shift $\Delta\lambda_R$ is negligible enough to not affect the value of SpO_2 significantly. From the above expression, it can be seen that any changes in the source LED wavelength can be corrected or accounted for in the SpO_2 calculation by choosing appropriate values of extinction coefficient for oxygenated and deoxygenated blood. In other words, if the value of $\Delta\lambda_R$ is known, its variation in the corresponding SpO_2 calculation can be accounted for and corrected accordingly.

In order to determine the value of $\Delta\lambda_R$, a calibration device having a waveguiding structure in conjunction with a filter in the form of an optical edge filter can be used. The optical waveguiding structure may be fabricated of a material which is substantially transparent to RED wavelength of about 660nm wavelength and it guides light emitted by LED to a photo-detector. In the example embodiment described herein, the RED LED is preferably chosen instead of the IR LED in a pulse oximeter as the error introduced in the SpO_2 calculation due to the variation in RED LED wavelength is significantly larger than that due to the IR LED. It will be appreciated that the described method and device may be similarly applicable to wavelength shifts in the IR LED if desired. In some exemplary embodiments, the light can be guided either by incorporating angled mirrors disposed at angles for example at about 45° to the emission plane of the light source or a channel waveguide with suitable coupling structures. Examples of suitable waveguides can be seen in Fig. 6, Fig. 7, Fig. 11 and Fig. 15. It would also be appreciated that the calibration device, waveguides and/or filter may exist separately or be incorporated in a PPG measuring device such as a pulse oximeter.

In some example embodiments, the shift in the wavelength of the source LED can be determined by passing its light through a filter, for example an optical edge filter which has a steep transmission change around the desired LED wavelength, λ_0 as shown in Fig.3. If the PPG device during manufacture has a certain wavelength for a specific driving current, the transmitted light through the optical filter is expected to be of a certain power. However, if the LED's wavelength changes, the power transmitted through the optical filter changes correspondingly. As such, this change in the transmitted light power can be used to determine the change in LED's wavelength, $\Delta\lambda$. An example of the transmission spectrum of an optical edge filter is shown in Fig. 3, and an example of the power spectrum of the light after transmitting through the optical edge filter is shown respectively in Fig. 4, with ΔP_{det} representing the change in detected power from the expected power. An example of the transmitted power through the optical edge filter as a function of wavelength of the LED is

shown in Fig. 5. In the qualitative description of this exemplary embodiment, it is assumed that the bandwidth of the LED does not change appreciably as the LED ages. Once the shift in the LED wavelength is obtained, the above equations (3) to (6) may be used to determine an accurate SpO_2 value.

Also due to thermal and mechanical conditions, the output power of the LED may vary over time for a constant driving current. This may result in a false reading of a change in wavelength. In order to counteract this, an unobstructed reference path is desired so that the detector can measure the power of the LED in the absence of the optical edge filter. In one example embodiment, this can be achieved by the presence of two or more optical waveguiding paths in the calibration device of which one acts as a reference path for detecting the total power emitted by the LED and the other path with an optical edge filter that measures the change in wavelength, $\Delta\lambda$, of the LED. Examples of suitable waveguides in the above described forms can be seen in Fig. 6, Fig. 7 and Fig. 15.

In an example embodiment, the two optical waveguiding paths can be incorporated in a single calibration device by orienting the two waveguides substantially perpendicular to each other such that they can be rotated about an axis perpendicular to the plane of the two waveguides. An example of a suitable calibration device having two optical waveguiding paths is provided in Fig, 6.

The calibration device of Fig. 6 comprises a light shielding cover 600 for shielding ambient light from the photodetector when used together with a PPG device. The calibration device comprises a single waveguiding structure 602 with a two waveguiding optical paths intersecting at about 90° with each other that allows light to travel from the light source to the light detector for e.g. a photodetector when in optical communication with the light source. The waveguiding structure 602 is disposed on a base 604. The calibration device also comprises a filter disk 606 in the form of an optical edge filter having substantially

transparent access holes e.g. 608 thereon. The substantially transparent access holes may be through holes or may be covered by a substantially transparent material. All parts of the filter disk 606 act to filter light except the parts where the access holes e.g. 608 are present.

Referring to Fig. 7a, when in use, the waveguiding structure 602 and the filter disk 606 may be rotated together such that in an unfiltered mode, the access holes e.g. 608 are substantially aligned with a light source 704 in the form of a LED and a light detector in the form of a photodetector 702 respectively, such that the light beam passing from the light source 704 through the access holes e.g. 608, travel along the waveguiding structure 602 before reaching the photodetector 702 unfiltered. In other words, the filtering part of the filter disk is configured to be disposed out of the path of light travelling from the light source 704 to the light detector 702 such that the light detected by the light detector is unfiltered light.,

Referring now to Fig. 7b, in a filtering mode, the waveguiding structure 602 and the filter disk 606 may be rotated together such that the filtering part of the filter disk 606 is disposed in the path of the light beam travelling from the light source 704 to the photodetector 702 to produce filtered light. In other words, the filtering part of the filter disk is configured to be disposed in the path of light travelling from the light source 704 to the light detector 702 such that the light detected by the light detector is filtered light.

As shown in Fig. 7a and Fig. 7b, the waveguiding structure 602 also comprises reflective surfaces or mirrors 610 disposed at an angle of about 45° to the base 604 to provide an optical path from the light source 704 to the photodetector 702.

In the orientation shown in Fig. 7a the reference power (of unfiltered light) is measured and in the 90° rotated orientation, the filtered output from the LED is measured as shown in Fig. 7b.

An exemplary flowchart (Fig. 14a) outlining the utilization of a wavelength calibration device having two mutually perpendicular waveguiding paths such as that shown in Fig, 6 and Figs. 7(a) and 7(b), is described in more detail below.

In the embodiment described above, the optical edge filter has a steep transmission spectrum around the wavelength of interest for the LED shown in Fig. 8. For example, if the LED has a peak wavelength of 650nm (λ_0) and a bandwidth of 20nm, then the optical edge filter is configured to have a very steep transmission change from about 600 to about 700nm (for example 10% transmission at 600nm and 90% transmission at 700nm). The radiation spectrum of the LED can be defined by approximating it to a lorentzian function shown in Fig. 8 and represented as:

$$P_{LED}(\lambda) = C \cdot \frac{\frac{1}{2} \cdot \Gamma}{(\lambda - \lambda_0)^2 + \left(\frac{1}{2} \cdot \Gamma\right)^2} \qquad (7)$$

Where, Γ is the bandwidth of the LED and λ_0 is its peak wavelength and C is the normalization constant. The total power emitted by the LED can then be written as:

$$P_{total} = \int_{\lambda} \int_{\Omega} P_{LED}(\lambda) d\lambda d\Omega \qquad (8)$$

Where, the above expression takes into account the angular distribution (Ω) of the emitted light from the LED. If this light is passed through an optical edge filter which has a transmission spectrum given by $T(\lambda)$, the transmitted light can be written as:

$$P_{det} = \iint_{\lambda,\Omega} P_{LED}(\lambda) \cdot T(\lambda) d\lambda d\Omega \qquad (9)$$

Depending upon the relation between the transmission spectrum of the optical edge filter and wavelength, the detected power, P_{det}, may change. A spectrum of the LED light after passing through the optical edge filter is shown in Fig. 4. The change in the spectrum shape from the original lorentzian shape is due to the steep transmission curve of the filter, for e.g. the optical edge filter.

The power falling on the detector after traversing through the waveguiding structures can be taken to be P_0, In such a case, the power, P_0, incident on the detector through the waveguiding structure without any optical edge filter may be just a fixed fraction of the total light emitted by the LED. This fixed fraction may depend on the radiation pattern of the LED and the geometrical arrangement of the 45° mirrors of the waveguiding structure with respect to the LED. This fraction, f, can be mathematically written as:

$$\begin{aligned} f &= \frac{P_0}{P_{total}} \\ P_{total} &= \iint_{\lambda,\Omega} P_{LED}(\lambda,\Omega)d\lambda d\Omega \\ P_0 &= f_1 \cdot \int_{\lambda} P_{LED}(\lambda)d\lambda \\ P_{det} &= f_2 \cdot \int_{\lambda} P_{LED}(\lambda)\cdot T(\lambda)d\lambda \end{aligned} \qquad (10)$$

However, as the radiation pattern of a given LED as well as the geometrical arrangement of the waveguiding structure is a constant, this fraction, f, can be safely taken out of the integral (equation (8)). In the above equation (10), f_1 and f_2 are the fractions corresponding to the two mutually perpendicular waveguiding paths. The ratio $K = P_{det}/P_0$ can then provide an accurate measure of the change in wavelength of the LED, $\Delta\lambda$ as the functional form of $T(\lambda)$ is known. Thus, the ratio of the power detected by the photo-detector in the presence and absence of the optical edge filter can be represented as:

$$\frac{P_{det}}{P_0} = \frac{f_2}{f_1}\frac{\int_{\lambda} T(\lambda)\cdot P_{LED}(\lambda)d\lambda}{\int_{\lambda} P_{LED}(\lambda)d\lambda} \qquad (11)$$

In the case of the embodiment having two waveguides substantially perpendicular to each other such that they can be rotated about an axis perpendicular to the plane of the two waveguides, for example as shown in Figs. 6, 7 and 15, the f_1 and f_2 values can be slightly different due to manufacturing of the two 45° angled mirror facets or the geometrical arrangement of the LED with respect to the mirrors. However, the f_1 and f_2 values can be approximated to be equal and can removed from equation (11).

In another alternate embodiment, where only single light guiding path is utilized in conjunction with a filter that can be interchangeably disposed in or out of the light path, for example as show in Fig. 11, the fractions f_1 and f_2 are essentially the same and can be factored out in equation (11).

The detected power P_{det}, can then be compared with the detected power at the time of manufacturing, P_{det0}, using the same measurement protocol as described above and detailed in a flowchart shown in Fig. 9. Any differences in these two detected power P_{det} and P_{det0} should fire a sequence of events shown in Fig. 10, which can account for a shift in the source LED's wavelength in the calculation of SpO_2 by the device. The steps shown in Fig. 9 and Fig. 10 are explained below.

In Fig 9, at step 902, the look-up table of the optical edge filter transmission vs. wavelength can be input into a memory of the PPG device. In step 904, the look-up table for ε_{Hb} and ε_{HbO2} or SpO_2 calibration curves for different wavelengths can also be input into the memory of the PPG device. Thereafter, in step 908, a calibration driving current for the LED, I_0 is set. The waveguiding structure of a calibration device, can then be placed above the LED and PD without the presence of the optical edge filter to record the incident power P_{i0}, as shown in step 910. In step 912, the calibration structure (or the filter disk) is rotated for example by 90° such that the optical edge filter is in the waveguiding path between the LED and PD. The detected power P_{det0} of the filtered light is then recorded. Subsequently, in step 914, the ratio $K_0=P_{det0}/P_{i0}$ is calculated. In step 916, the threshold value of $\Delta\lambda_{th}$ for which SpO_2 should be corrected is then input into the memory of the PPG device. In step 918, the values of K_0, $\Delta\lambda_{th}$, and I_0 are stored in the memory. Step 920 marks the end of the measurement protocol at the time of manufacturing the PPG device.

In Fig. 10, step 1002 marks the start of the calibration routine. In step 1004, the change in wavelength $\Delta\lambda$ is measured. The modulus of $\Delta\lambda$ is obtained to determine if the numerical value of $\Delta\lambda$ is greater than the threshold wavelength change $\Delta\lambda_{th}$. If affirmative, the calibration routine proceeds to step 1008 to change the SpO_2 expression corresponding to $\Delta\lambda$. If negative, the calibration routine proceeds to end at step 1010.

Another embodiment of the calibration device that was briefly mentioned above utilizes a single light guiding path that incorporates a movable filter disk placed in between the reflective PPG device and the optical waveguide path which measures the power transmitted by the source LED in the presence and absence of the optical edge. This can alleviate the manufacturing inaccuracies that could occur in the fabrication of the mirror facets of the waveguiding structures. Imperfections in the fabrication of the mirror facets could result in different amounts of light being detected in the absence of the optical edge filter for the two optical waveguiding paths. This error could result in false readings for the wavelength calibration device, thus, compounding the errors introduced in the measurement of SpO_2. If only a single light path is used, then light detected by the photo-detector stays constant in the absence of the optical edge filter and placing the filter in the light path provides a better indication of the shift in the wavelength of the source LED. One example embodiment of a calibration device which utilizes a single light guiding path is provided in Fig, 11, where an exploded view of the calibration device is shown.

The calibration device of Fig. 11 comprises a light shielding cover 1100 for shielding ambient light from the photodetector PD when used together with a PPG device. The calibration device comprises a single waveguiding structure 1102 with a single path that allows light to travel from the light source to the photodetector when in optical communication with the light source. The waveguiding structure 1102 is disposed on a base 1104. The calibration device also comprises a filter disk 1106 having a filter patch 1108 thereon. All parts of the filter disk 1106 are substantially transparent to light except the part

where the filter patch 1108 is present. The calibration device can also comprise a clear supporting disk 1110. The calibration device can also comprise reflective surfaces or mirrors disposed at an angle of about 45° to the base 1100 to provide an optical path from a light source to a photodetector. These reflective surfaces or mirrors are represented in Fig. 11 as sloping surfaces.

When in use, the filter disk 1106 may be rotated such that in a filtering mode, the filter patch 1108 is substantially aligned with the light source to produce a filtered light beam which travels along the waveguiding structure 1102 towards the photodetector. In other words, in the filter mode, the filter patch 1108 of the filter disk is configured to be disposed in the path of light travelling from the light source to the light detector such that the light detected by the light detector is filtered light.

In an unfiltered mode, the filter disk 1106 may be rotated such that in the filter patch 1108 is misaligned with the light source such that the light beam passing from the light source through the clear supporting disk 1110, the transparent parts of the filter disk 1106 , along the waveguiding structure 1102 before reaching the photodetector is substantially unfiltered. In other words, in the unfiltered mode, the filter patch 1108 of the filter disk is configured to be disposed out of the path of light travelling from the light source to the light detector such that the light detected by the light detector is unfiltered light.

An exemplary flowchart (Fig. 14b) outlining the utilization a wavelength calibration device having a single waveguiding path and a rotatable optical filter disk such as that shown in Fig, 11, is described in more detail further below.

There are two exemplary ways to account for the change in wavelength of the LED for the calculation of SpO_2 in a pulse oximeter.

One way is to have a look-up table for different values of the extinction coefficients of oxygenated and de-oxygenated hemoglobin at finite wavelength intervals (for example, 1nm wavelength interval) and plug-in/input the appropriate value corresponding to the measured wavelength change in the SpO_2 calculation expression to obtain a corrected SpO_2 result when the measured change in wavelength of the LED is greater than the specified wavelength interval, $\Delta\lambda_{th}$. This is shown in Fig. 13, where step 1300 marks the start of the calibration routine. In step 1302, the change in wavelength $\Delta\lambda$ is measured. The change in wavelength $\Delta\lambda$ is determined whether to be greater than the threshold wavelength change $\Delta\lambda_{th}$ in step 1304. If affirmative, the calibration routine proceeds to step 1306 to change the SpO_2 expression corresponding to $\Delta\lambda$ by using ε_{Hb} and ε_{HbO2} values corresponding to $\lambda_R = \lambda_R + \Delta\lambda$. If negative, the calibration routine proceeds to end at step 1308.

A second way of compensating for the change in wavelength of the LED is to have pre-defined calibration curves relating SpO_2 and R values for different wavelengths of the LED at the time of manufacture. These different wavelengths can be periodically spaced, at regular intervals (for example, at wavelength spacing of 1nm). Thus, if the calibration device measures a wavelength change exceeding the specified wavelength interval, a new look-up table can be programmed into the device for measuring the SpO_2 value. This is shown in Fig. 12, where step 1200 marks the start of the calibration routine. In step 1202, the change in wavelength $\Delta\lambda$ is measured. The change in wavelength $\Delta\lambda$ is determined whether to be greater than the threshold wavelength change $\Delta\lambda_{th}$ in step 1204. If affirmative, the calibration routine proceeds to step 1206 to change the SpO_2 expression corresponding to $\Delta\lambda$ by looking up a table corresponding to SpO_2 variation with R for $\lambda_R = \lambda_R + \Delta\lambda$. If negative, the calibration routine proceeds to end at step 1208.

Fig. 14a is a flowchart showing the utilization of two substantially mutually perpendicular waveguiding paths of a wavelength calibration device in accordance with an

exemplary embodiment disclosed herein. An example of a calibration device having two substantially mutually perpendicular waveguiding paths is shown in Fig. 6.

Step 1400 of Fig. 14a marks the start of the calibration routine. In step 1402, the calibration device is placed on top of a light source for e.g. a LED and a light detector, for e.g. a photo-detector of a pulse oximeter and a waveguide is positioned without any optical edge filter. The power P_0 of the unfiltered light from the LED is then measured in step 1404. In step 1406, the calibration device is rotated by 90° and the waveguide containing a filter in the form of an optical edge filter is placed in the path between the LED and the photo-detector of the pulse oximeter. The power P_{det} of the filtered light from the LED is then measured in step 1408. In step 1410, the ratio P_{det}/P_0 is then calculated and the value of the wavelength change $\Delta\lambda$ is obtained from a look-up table for the transmittance of the optical edge filter as a function of wavelength.

Fig. 14b is a flowchart showing the utilization of a single waveguiding path and a rotatable optical filter disk of a calibration device in accordance with an exemplary embodiment disclosed herein. An example of a calibration device having a single waveguiding path and a rotatable optical filter disk is shown in Fig. 11.

Step 1500 of Fig. 14b marks the start of the calibration routine. In step 1502, the calibration device is placed on top of a light source in the form of an LED and a light detector in the form of an photo-detector (PD) of a pulse oximeter and a waveguide is positioned with the clear/transparent part of the filter disk between the LED and PD. The power P_0 of the unfiltered light from the LED is then measured in step 1504. In step 1506, the filter disk is rotated such that a filter in the form of an optical edge filter is placed in the path between the LED and the photo-detector of the pulse oximeter. The power P_{det} of the filtered light from the LED is then measured in step 1508. In step 1510, the ratio P_{det}/P_0 is then calculated and

the value of the wavelength change $\Delta\lambda$ is obtained from a look-up table for the transmittance of the optical edge filter as a function of wavelength.

Embodiments of the method and calibration device disclosed herein, advantageously allow for the compensation of discrepancies or inaccuracies caused by the variation of the wavelength of the light source of physiological measurement devices such as oximeters during the measurement of SpO_2. Accordingly, embodiments of the disclosed method and calibration device allows the initial accuracy of oximeters to be maintained even after prolong use. Even more advantageously, as embodiments of the disclosed method may be carried out using only a single light source and a single light detector, the method is an effective yet simple way of obtaining a correction factor to compensate for wavelength shifts of the light source. Furthermore, embodiments of the disclosed method require detection of only two optical characteristics P_0 and P_{det} to obtain the correction factor.

Embodiments of the disclosed calibration device may advantageously be coupled or decoupled from the physiological measurement devices. Accordingly, a single calibration device may be used to calibrate a plurality of physiological measurement devices, such as oximeters. Embodiments of the calibration device also provide for an unobstructed reference light path for measurement of the power of the light source in the absence of a filter e.g. optical edge filter to obtain a reference point for the output power of the light source. Such reference point further serves to improve the reliability of the correction factor obtained, leading to more accurate compensated SpO_2 readings obtainable for calibrated oximeters. Advantageously, embodiments of the calibration devices utilize mirrors to reduce complicacy of the light paths provided by the waveguiding structures. Certain embodiments of the calibration devices disclosed herein also comprises a single path for allowing filtered and unfiltered light to travel from the light source to the light detector. This advantageously reduces the likelihood of discrepancies/inaccuracies of the obtained correction factor caused

by physical aberrations (such as distance etc) or dissimilarities between the light path for filtered light and unfiltered light. Even more advantageously, embodiments of the calibration device are compact and easily portable for ease of use.

It will be appreciated by a person skilled in the art that other variations and/or modifications may be made to the specific embodiments without departing from the spirit or scope of the invention as broadly described. The present embodiments are, therefore, to be considered in all respects to be illustrative and not restrictive.

CLAIMS

1. A method of obtaining a correction factor to compensate for wavelength shifts of a light source, the method comprising:
   (i) detecting by a light detector, a first optical characteristic of substantially unfiltered light emitted by the light source;
   (ii) detecting by the light detector, a second optical characteristic of filtered light from the light source; and
   (iii) comparing the first optical characteristic, the second optical characteristic and a predetermined set of values to obtain the correction factor.

2. The method as claimed in claim 1 wherein the light source and the light detector are comprised in a physiological parameter measurement device.

3. The method as claimed in claims 1 or 2, wherein the comparing step (iii) comprises obtaining a ratio of the first optical characteristic to the second optical characteristic.

4. The method as claimed in claim 3 , wherein the comparing step (iii) further comprises:
   comparing the ratio with a predetermined ratio to obtain a value of the wavelength shift of the light source; and
   corresponding the value of the wavelength shift with a table of predetermined values to obtain the correction factor.

5. The method as claimed in claim 4, wherein the predetermined values of the table of predetermined values are provided at incremental wavelengths.

6. The method as claimed in any one of claims 1 to 5, wherein the predetermined ratio is a ratio of the first optical characteristic to the second optical characteristic obtained at the time of manufacturing the light source by performing steps (i) to (ii).

7. The method as claimed in claim 2, wherein the predetermined ratio is a ratio of the first optical characteristic to the second optical characteristic obtained at

the time of manufacturing the physiological parameter measurement device by performing steps (i) to (ii).

8. The method as claimed in any one of claims 1 to 7, wherein the first and second optical characteristics are values of the power of the light detected by the light detector.

9. The method as claimed in any one of claims 1 to 8, wherein the filtered light is produced by passing light from the light source through a filter.

10. The method as claimed in claim 8, wherein the filter is an optical edge filter.

11. The method as claimed in any one of claims 1 to 9, wherein the physiological parameter measurement device comprises a photoplethysmography (PPG) sensing device.

12. The method as claimed in claim 11, wherein the photoplethysmography (PPG) sensing device comprises a pulse oximeter.

13. The method as claimed in any one of the preceding claims, wherein the light detector comprises a photodetector.

14. The method as claimed in any one of the preceding claims, wherein the light is selected from the group consisting of X-rays, visible light rays, ultraviolet light rays and infra-red rays.

15. The method as claimed in claim 3, wherein the ratio is represented by the formula

$$\frac{P_{det}}{P_0} = \frac{f_2}{f_1} \frac{\int_\lambda T(\lambda) \cdot P_{LED}(\lambda) d\lambda}{\int_\lambda P_{LED}(\lambda) d\lambda},$$

wherein P_0 represents the power of the filtered light detected by the light detector, P_{det} represents the power of the substantially unfiltered light detected by the light detector, $T(\lambda)$ represents the transmission spectrum of the light source as a function of the wavelength λ, $P_{LED}(\lambda)$ represents the radiation spectrum of the light source as a function of the wavelength λ, and f_1

and f_2 represent constants which are based on the light path of the substantially unfiltered and filtered light respectively.

16. The method as claimed in claim 6, wherein the predetermined ratio is represented by the formula

$$\frac{P_{det0}}{P_{i0}} = \frac{f_2}{f_1} \frac{\int_{\lambda} T(\lambda) \cdot P_{LED}(\lambda) d\lambda}{\int_{\lambda} P_{LED}(\lambda) d\lambda},$$

wherein P_{i0} represents the power of the filtered light detected by the light detector at the time of manufacturing the light source, P_{det0} represents the power of the substantially unfiltered light detected by the light detector at the time of manufacturing the light source, $T(\lambda)$ represents the transmission spectrum of the light source as a function of the wavelength λ, $P_{LED}(\lambda)$ represents the radiation spectrum of the light source as a function of the wavelength λ, and f_1 and f_2 represent constants which are based on the light path of the substantially unfiltered and filtered light respectively.

17. The method as claimed in claim 7, wherein the predetermined ratio is represented by the formula

$$\frac{P_{det0}}{P_{i0}} = \frac{f_2}{f_1} \frac{\int_{\lambda} T(\lambda) \cdot P_{LED}(\lambda) d\lambda}{\int_{\lambda} P_{LED}(\lambda) d\lambda},$$

wherein P_{i0} represents the power of the filtered light detected by the light detector at the time of manufacturing the physiological parameter measurement device, P_{det0} represents the power of the substantially unfiltered light detected by the light detector at the time of manufacturing the physiological parameter measurement device, $T(\lambda)$ represents the transmission spectrum of the light source as a function of the wavelength λ, $P_{LED}(\lambda)$ represents the radiation spectrum of the light source as a function of

the wavelength λ, and f_1 and f_2 represent constants which are based on the light path of the substantially unfiltered and filtered light respectively.

18. A calibration device for coupling to a physiological measurement device having a light source and a light detector, the calibration device comprising:

a wave guiding structure configured to allow light to travel from the light source to the light detector along a predetermined path, when the calibration device is coupled to said physiological measurement device; and

a filter for producing filtered light from the light source when the calibration device is coupled to said physiological measurement device, the filter configured to be disposed in or out of the path of light travelling from the light source to the light detector,

wherein the calibration device allows the physiological measurement device to perform the method as claimed in any one of claims 1 to 13 when coupled thereto.

19. The calibration device as claimed in claim 18, wherein the wave guiding structure comprises at least two separate paths, each of said separate path allowing light to travel from the light source to the light detector when in optical communication with the light source.

20. The calibration device as claimed in claim 19, wherein the at least two separate paths are configured to be in interchangeable optical communication with the light source.

21. The calibration device as claimed in any one of claims 18 to 20, wherein the wave guiding structure comprises a plurality of mirrors.

22. The calibration device as claimed in claim 21, wherein a reflective plane of at least one mirror is positioned at angle of about 45° to an emission plane of the light source and a reflective plane of at least another mirror is positioned at an angle of about 45° to an incident plane of the light detector, the emission plane being substantially perpendicular to the direction of the emitted light from the light source and the incident plane being substantially perpendicular to the direction of light incident on the light detector.

23. The calibration device as claimed in claim 20, wherein the at least two separate paths intersect with each other at an angle of about 90°.

24. The calibration device as claimed in any one of claims 18 to 23, further comprising a light shielding cover for shielding ambient light from the light detector when the calibration device is coupled to the physiological measurement device.

25. The calibration device as claimed in any one of claims 18 to 24, wherein the filter is disposed on a rotatable disc.

26. The calibration device as claimed in any one of claims 18 to 25, wherein the wave guiding structure is disposed on a rotatable disc.

27. The calibration device as claimed in any one of claims 18 to 26, wherein the filter is an optical edge filter.

28. The calibration device as claimed in any one of claims 18 to 27, wherein the physiological parameter measurement device comprises a photoplethysmography (PPG) sensing device.

29. The calibration device as claimed in claim 28, wherein the photoplethysmography (PPG) sensing device comprises a pulse oximeter.

30. The calibration device as claimed in any one of claims 18 to 29, wherein the light detector comprises a photodetector.

31. The calibration device as claimed in any one of claims 18 to 30, wherein the light is selected from the group consisting of X-rays, visible light rays, ultraviolet light rays and infra-red rays.

32. A physiological parameter measurement device for performing the method as claimed in any one of claims 1 to 17, the measurement device comprising:

a light source for emitting light;

a light detector for measuring an optical characteristic of light emitted by the light source; and

a filter for producing filtered light from the light source, the filter configured to be interchangeably disposed in or out of the path of light travelling from the light source to the light detector.