R E V I E W

Pulse oximetry: fundamentals and technology update

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Abstract: Oxygen saturation in the arterial blood (SaO_2) provides information on the adequacy of respiratory function. SaO_2 can be assessed noninvasively by pulse oximetry, which is based on photoplethysmographic pulses in two wavelengths, generally in the red and infrared regions. The calibration of the measured photoplethysmographic signals is performed empirically for each type of commercial pulse-oximeter sensor, utilizing in vitro measurement of SaO_2 in extracted arterial blood by means of co-oximetry. Due to the discrepancy between the measurement of SaO_2 by pulse oximetry and the invasive technique, the former is denoted as SpO_2. Manufacturers of pulse oximeters generally claim an accuracy of 2%, evaluated by the standard deviation (SD) of the differences between SpO_2 and SaO_2, measured simultaneously in healthy subjects. However, an SD of 2% reflects an expected error of 4% (two SDs) or more in 5% of the examinations, which is in accordance with an error of 3%–4%, reported in clinical studies. This level of accuracy is sufficient for the detection of a significant decline in respiratory function in patients, and pulse oximetry has been accepted as a reliable technique for that purpose. The accuracy of SpO_2 measurement is insufficient in several situations, such as critically ill patients receiving supplemental oxygen, and can be hazardous if it leads to elevated values of oxygen partial pressure in blood. In particular, preterm newborns are vulnerable to retinopathy of prematurity induced by high oxygen concentration in the blood. The low accuracy of SpO_2 measurement in critically ill patients and newborns can be attributed to the empirical calibration process, which is performed on healthy volunteers. Other limitations of pulse oximetry include the presence of dyshemoglobins, which has been addressed by multiwavelength pulse oximetry, as well as low perfusion and motion artifacts that are partially rectified by sophisticated algorithms and also by reflection pulse oximetry.

Keywords: oxygen saturation, pulse oximetry, photoplethysmography, arterial blood, venous blood

Arterial oxygen saturation

The transfer of oxygen from the lungs to the tissue cells is carried out mainly by the hemoglobin molecules in the red blood cells. The total oxygen content in blood includes the hemoglobin-bound oxygen (97%–98% of the total oxygen content) and the oxygen dissolved in plasma. The level of arterial hemoglobin oxygenation is assessed by oxygen saturation in arterial blood (SaO_2), which is the ratio of oxygenated hemoglobin concentration [HbO_2] to total hemoglobin concentration in the blood ([HbO_2] + [Hb]):

$$SaO_2 = [HbO_2]/([HbO_2] + [Hb]). \qquad (1)$$

SaO_2 has the same value throughout the arterial system, since oxygen is extracted from the blood only in the capillaries. The concentration of dissolved oxygen in arterial

blood is measured by arterial oxygen partial pressure (PaO_2). SaO_2 increases as PaO_2 increases in an S-shaped curve, the dissociation curve, which depends on blood temperature, acidity level, and the concentration of several substances in the blood. Typical values of PaO_2 for adults at sea level range between 80 and 100 mmHg and those of SaO_2 between 96% and 98%. Because of the gradual slope of the upper part of the dissociation curve, a change of PaO_2 from 100 to 70 mmHg under normal conditions only results in a decrease of SaO_2 from 97% to 92%. With regard to venous blood, the normal range of oxygen saturation is 70%–80%, and oxygen partial pressure varies in the range of 40–50 mmHg.[1,2]

PaO$_2$ and SaO$_2$ have major clinical and physiological significance, since they are dependent on the adequacy of respiratory function and are directly related to the oxygen supply to the organs. Both PaO_2 and SaO_2 can be obtained from a sample of extracted arterial blood: PaO_2 can be measured with an arterial blood gas analyzer and SaO_2 by co-oximetry, which uses the different light absorption spectra for oxygenated and deoxygenated hemoglobin. PaO_2 and SaO_2 can also be measured noninvasively. The noninvasive transcutaneous PaO_2 electrode has low accuracy and requires heating of the skin to 43°C–44°C.[3,4] The noninvasive technique of pulse oximetry[5–7] for the assessment of SaO_2 is the subject of the current review. After describing the fundamentals of the technique, the review discusses the origins and the level of inaccuracy in oxygen-saturation measurement by pulse oximetry, as well as the clinical significance of the error in SaO_2 measurement, particularly in newborns.

Pulse oximetry – the technique

The optical techniques that have been developed for the assessment of SaO_2 are based on the different light-absorption spectra for HbO_2 and Hb. Figure 1 shows the extinction coefficients – the specific absorption constants – of HbO_2 and Hb as a function of wavelength in the visible and near-infrared regions. The extinction coefficient of each type of hemoglobin is defined as the absorption constant of the hemoglobin in a sample, divided by the hemoglobin concentration in the sample. The hemoglobin in blood includes HbO_2 of extinction coefficient ε_O and Hb of extinction coefficient ε_D, and the total extinction coefficient in the arterial blood, ε, is related to its SaO_2 by:

$$\varepsilon = \varepsilon_O\, SaO_2 + \varepsilon_D\, (1 - SaO_2), \qquad (2)$$

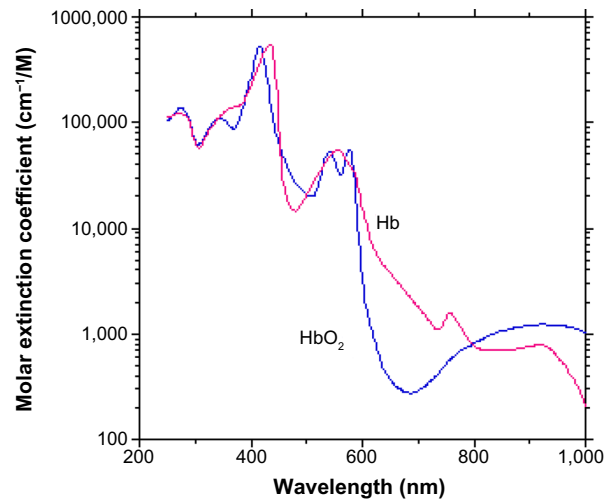so that light-absorption measurements can provide assessment of SaO_2.

Figure 1 Absorption spectra of the oxygenated and deoxygenated hemoglobin molecules.
Notes: In the red and the infrared regions, the absorption is relatively low and allows accurate measurement of light transmission. Copyright © 1999. Prahl S. Reproduced from Prahl S. Optical absorption of hemoglobin. 1999. Available from: http://omlc.ogi.edu/spectra/hemoglobin/index.html. Accessed May 26, 2014.[8]
Abbreviations: HbO$_2$, oxygenated hemoglobin; Hb, deoxygenated hemoglobin.

Hemoglobin is the main source for light absorption in tissue in the red and near-infrared regions, but other chromophores like melanin and myoglobin can also absorb light in these regions. Venous blood, with less oxygenated hemoglobin, also absorbs light in the same spectral region as that of arterial blood. The need to isolate the contribution of hemoglobin in arterial blood to total absorption has led to the development of pulse oximetry.

Pulse oximetry for the assessment of SaO_2 is based on photoplethysmography (PPG), the measurement of light-absorption increase due to the systolic increase in arterial blood volume.[5,6] The PPG signal is shown in Figure 2. Transmitted light intensity decreases during systole, when blood is ejected from the left ventricle into the vascular system, thereby increasing the peripheral arterial blood volume. The maximal and minimal values of the PPG pulse reflect
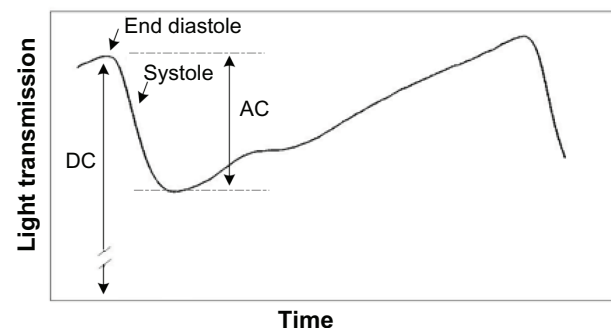

Figure 2 The photoplethysmography signal.
Note: DC denotes the pulse baseline and AC the pulse amplitude.

light irradiance transmitted through tissue when tissue blood volume is minimal or maximal, respectively. The PPG amplitude is related to the light absorption in the arterial blood volume increase during systole.

The technique of pulse oximetry has been described in several publications.[6,9–11] PPG measurement in each wavelength enables the assessment of the contribution of arterial blood to the total absorption of light, assuming that the PPG signal reflects changes in arterial blood volume. The PPG-signal amplitude (generally denoted AC) divided by its baseline (generally denoted DC) is related to the maximal blood volume change during systole.[12] In order to measure SaO_2, PPG curves in two wavelengths are recorded, and SaO_2 is derived from the ratio of ratios, R, defined as:

$$R = \frac{(AC/DC)_1}{(AC/DC)_2} \quad (3)$$

Direct determination of SaO_2 from PPG measurements in several wavelengths using the Beer–Lambert law is not applicable, because light scattering in tissue and blood also affects attenuation of light in tissue. Light scattering in blood is due to the difference in the refractive index between red blood cells and plasma, and light scattering in tissue is attributed to the difference in refractive index between cellular organelles and cellular fluid, as well as between intracellular and extracellular fluids.[13,14] Scattering results in the escape of light from tissue in various directions, and also increases the path length of light in tissue, thereby increasing the probability for absorption in the blood. In order to determine the value of SaO_2 for blood in tissue from light-transmission measurements, the contribution of light absorption to the total attenuation must be isolated.

In commercial pulse oximeters, the two wavelengths are chosen in the red and infrared regions, where the difference in light absorption between the two wavelengths is relatively large. However, the scattering constant and the optical path length differ significantly between the red and infrared wavelengths, and consequently the relationship between the physiological parameter SaO_2 and the measured parameter R cannot be derived directly from physical and physiological considerations of light absorption in HbO_2 and Hb, based on the Beer–Lambert law. The relationship between R and SaO_2 is determined experimentally for each type of commercial pulse oximeter sensor by calibration:[6,7] R is measured in several healthy volunteers simultaneously with in vitro measurement of SaO_2 in extracted arterial blood by means of

co-oximetry. The formula relating R to SaO_2 is determined by proposing a mathematical relationship, such as:

$$SaO_2 = \frac{k_1 - k_2 R}{k_3 - k_4 R} \quad (4)$$

and obtaining the values of the constant k_i for the specific pulse oximeter by best-fit analysis of the measured parameters in the calibration process.

Empirical calibration is based on the assumption that the relationship between the measured parameter R and the physiological parameter SaO_2 is not influenced by intersubject variability in the circulatory system. However, a change in the optical path length, if not equivalent in the red and infrared wavelengths, can change the relationship between R and SaO_2.[15,16] If the red–infrared path-length ratio changes between different subjects, in particular between the healthy subjects on whose fingers the empirical calibration was performed and the patients on whose fingers the clinical examination was carried out, inaccuracy in SaO_2 measurement could be expected. The SaO_2 value measured by pulse oximetry is denoted as SpO_2, and its deviations from the SaO_2 value directly measured in extracted blood are discussed.

The accuracy of pulse oximetry

The accuracy of a pulse oximeter is evaluated by the differences between SpO_2, the oxygen-saturation values measured by the pulse oximeter, and SaO_2, measured by co-oximetry in extracted blood, the gold standard.[17] Most manufacturers of pulse oximeters claim an accuracy of 2%, which is the standard deviation (SD) of the differences between SpO_2 and SaO_2. A standard deviation of 2% is associated with an expected error of 4% (two SDs) or more among 5% of the examinations (assuming that the distribution curve of the differences between SpO_2 and SaO_2 has normal distribution, the area under the curve at a distance greater than two SDs from the mean is 5% of the total area). In clinical studies, it was found that the accuracy for a single measurement of SpO_2 is 3%–4% and for monitoring SpO_2 in a specific patient 2%–3%.[17,18] Considering the fact that the relevant clinical range of SaO_2, including most sick patients, is 80%–100%, an error of 3%–4% could be of major significance. Despite this low accuracy, pulse oximetry enables the detection of an abrupt drop of SpO_2 by 3%–4% in patients during anesthesia or in an intensive care unit. It is accepted that a significant decrease in SpO_2 value obtained by the available commercial pulse oximeters is a reliable parameter for the detection of significant deterioration in respiratory function.

It should be known, however, that the accuracy of SpO_2 measurement is not equivalent to that of invasive SaO_2 measurement. In intensive care units, where inadequate oxygen supply to vital organs may be particularly harmful, maintaining a minimum SpO_2 level of 94% or 96% in mechanically ventilated patients has been proposed, in order to ensure a minimal SaO_2 value of 90%.[19,20] In a study on critically ill patients[21] the correlation between spontaneous changes in SpO_2 and in SaO_2 was found to be relatively low (r=0.6, r^2=0.37), leading the authors to conclude that changes in SpO_2 do not reliably predict equivalent changes in SaO_2 in the critically ill.

As was explained earlier, inaccuracy in SpO_2 measurement in critically ill patients is to be expected, because the empirical calibration of pulse oximeters is based on examinations on healthy volunteers and is not necessarily applicable to critically ill patients. The discrepancy between healthy volunteers examined during the empirical calibration process and patients is further accentuated in neonates.[22] The deviation of SpO_2 from SaO_2 is even greater at saturations below 70%–80%,[11,23–26] because ethical restrictions prevent manufacturers from reducing SaO_2 below 80% during the calibration process. The inaccuracy associated with the co-oximetry itself (upon which the calibration process is based) is an additional contributing factor to the error in SpO_2 measurement.[17]

The accuracy of SpO_2 measurement can be of great significance for critically ill patients undergoing oxygen therapy. Lately, evidence has accumulated to support the need for precise control of arterial oxygenation in order to avoid hyperoxemia and the ill effects of oxygen toxicity associated with it.[27] The detection of hyperoxemia in these patients is especially problematic, because the dissociation curve is almost flat in the high SaO_2 range, ie, greater than 95%, and thus relatively small changes in SaO_2 are associated with large changes in PaO_2. The limited ability of pulse oximetry to accurately determine the level of excess oxygenation is particularly important for preterm newborns receiving supplemental oxygen, due to their vulnerability to retinopathy of prematurity, induced by high PaO_2 in arterial blood. The significance of accurate SpO_2 measurement was demonstrated in three studies – SUPPORT (Surfactant, Positive Pressure, and Oxygenation Randomized Trial), BOOST (Benefits Of Oxygen Saturation Targeting) II, and COT (Canadian Oxygen Trial) – where 4,911 preterm newborns receiving oxygen supplementation were randomized to either a low (85%–89%) or high (91%–95%) SpO_2 value.[28] Increased risk of mortality was noted in the first group, while an increased incidence of retinopathy of prematurity was found in the second group. The authors of the meta-analysis

of those studies[28] recommended that SpO_2 should be targeted at 90%–95% in infants with gestational age <28 weeks. Some authors[29–31] suggest that pulse oximetry should not be the sole means for monitoring oxygenation in the neonatal intensive care unit.

An additional source for inaccuracy in SpO_2 measurement in newborns is fetal hemoglobin, which can constitute 95% of total hemoglobin and is slightly different from that of adult hemoglobin. The maximal expected error due to fetal hemoglobin in neonates was estimated by Mendelson and Kent[32] to be 3% (using theoretical simulations), and this error should be added to other sources of inaccuracy in SpO_2 measurement in neonates. Experimental examinations showed a 4% effect of fetal hemoglobin on neonatal pulse oximetry.[33,34]

Pulse oximetry for the detection of congenital heart diseases in neonates

Pulse oximetry has also been proposed as a newborn-screening test for the detection of critical congenital heart disease (CCHD), defined as CHD requiring surgery or catheter intervention in the first year of life.[35] This application of pulse oximetry is distinctive, because it provides an assessment of cardiac physiology, while the usual aim of SpO_2 measurement is the evaluation of respiratory function. Early detection of neonates with ductal-dependent CCHD is important, because their survival depends on the patency of the ductus arteriosus to ensure adequate pulmonary and systemic blood flow. Since the majority of infants with CCHD have some degree of hypoxemia during the newborn period,[35] pulse oximetry has been recommended as a screening test for the detection of neonatal CCHD before discharge, prior to the onset of symptoms. Though postnatal echocardiography is well established as the gold standard for diagnosing congenital heart diseases, it has significant limitations as a screening tool, mainly because of its cost and lack of availability of trained personnel to perform the examinations.[36]

The effectiveness of pulse-oximetry screening has been demonstrated in multiple international clinical trials. In a study in the UK,[36] more than 20,000 neonates were examined in the right hand and either foot. Saturation of <95% in either limb or a difference of >2% between the limb readings was taken as abnormal. In this study, pulse oximetry had a sensitivity of 75% for critical cases and a specificity of 99.16%. Similar criteria were suggested by the American Academy of Pediatrics:[37] saturation of ≥95% in either limb with a difference of ≤3% between the upper and lower limbs was taken as normal. In a systematic review and meta-analysis,[38] the authors selected studies that assessed the accuracy of

pulse oximetry for the detection of CCHD in asymptomatic newborns, and found high specificity (99.9%) with moderate sensitivity (76.5%).

Limitations of pulse oximetry and technological update
Dyshemoglobins and multiwavelength pulse oximetry

Apart from HbO_2 and Hb, adult blood may contain dyshemoglobin: hemoglobin derivatives, which are not functional because they are not able to reversibly bind oxygen molecules at physiological levels of PaO_2 in blood. The most important dyshemoglobins are methemoglobin (MetHb) and carboxyhemoglobin (COHb), which are commonly present in low concentrations in normal subjects. Increased concentration of dyshemoglobin molecules in blood (such as in CO poisoning) can reduce the effectiveness of tissue oxygenation. Functional SaO_2 is defined as the percentage of HbO_2 relative to the sum of HbO_2 and Hb, while fractional SaO_2 is defined as the percentage of HbO_2 relative to the total of four variants of hemoglobin.[39] At low concentrations of dyshemoglobins, the distinction between these two parameters is of negligible significance because of the small difference between them; at high-enough levels, both functional and fractional readings can be compromised.[30,40] A change of COHb concentration by 1% changes the pulse-oximetry reading by about 1%.[22]

Conventional pulse oximeters that utilize two wavelengths of light for the assessment of oxygen saturation are based on the assumption that HbO_2 and Hb are the only absorbers of light in these two wavelengths in the blood. Since MetHb and COHb absorb light in the wavelengths used in pulse oximetry,[22,40] an error in SpO_2 measurement is expected in the presence of these dyshemoglobins. Some manufacturers have developed pulse oximeters that use more than two light wavelengths, thereby enabling estimation of blood levels of COHb and MetHb (as well as total hemoglobin concentration). The accuracy of these measurements has been studied in healthy volunteers and among patients with suspected CO poisoning in emergency departments. Some studies showed accurate measurement of COHb and MetHb,[22,40–42] while others[43,44] claim that pulse co-oximetry cannot replace standard blood COHb measurement, though it could be used as a first-line screening test.

Low perfusion and reflection pulse oximetry

In transmission pulse oximetry, light is detected after being transmitted through an organ, and is therefore limited to fingertips and earlobes. The blood flow to the fingertips and earlobes is greater than what is required by tissue metabolism, due to their role in heat transfer, and under normal conditions their PPG pulses have a high signal-to-noise ratio. However, these organs are under intensive regulation by the autonomic nervous system, and in cases of low surrounding temperature or low cardiac output, their arteries are constricted in order to reduce heat dissipation or to maintain sufficient blood supply to the critical core organs: the heart, brain, and kidneys. In such cases, the PPG signal decreases, reducing pulse-oximeter accuracy. Reflection pulse oximetry, in which the light sources and the photodetector are located on the same surface of the skin, can be applied on any accessible site, and is thus of advantage in low peripheral perfusion conditions.[45,46]

The main site used for reflection pulse-oximetry measurement is the forehead. Studies in which a forehead sensor and a digit sensor were compared to SaO_2 measurements by co-oximetry showed conflicting results. In measurements on well-perfused pediatric patients, the forehead sensor was found to be as accurate as the digit sensor.[47] Comparison of forehead and digit sensors in critically ill surgical/trauma patients at risk for decreased peripheral perfusion showed lower bias between SpO_2 and SaO_2 for the forehead sensor,[48] and similar results were found in patients with low cardiac index.[49] Contradictory results showing inferior accuracy of reflective oximetry were found in a study on adults with acute respiratory distress syndrome during a high positive end-expiratory pressure recruitment maneuver.[50]

Some companies suggest reflection pulse oximeters for the finger. The advantage of reflection finger-pulse oximeters is their low power consumption, since the distance between the light sources and the detector can be shortened, resulting in lower light absorption. Reflection pulse oximeters were also suggested in accessible internal structures, such as the esophagus,[11,51,52] pharynx, and trachea.[53,54] Researchers claim that measurements at these sites are more reliable in conditions of low peripheral perfusion.

Low perfusion induced by vasoconstriction, which results in a decreased PPG signal, is also associated with an increase in SpO_2 value.[55–57] Local hyperthermia resulted in a significant decrease in SpO_2, while during local hypothermia SpO_2 increased.[56] A similar effect was found following administration of propofol/nitrous oxide anesthesia, leading to alteration of peripheral vascular tone and concomitant changes in skin temperature.[55] The observed increase in SpO_2 probably reflects decreased transmission of arterial pulsations to venous blood in the finger,[55] but it can also be speculated

that the effect is related to the calibration process. Changes in scattering parameters due to changes in microcirculation can interfere with the relationship between the measured parameter R and SaO_2 (Equation 4), which was obtained in healthy subjects under normal thermal conditions.

Calibration

In a former section we described the empirical calibration required for the determination of the relationship between R and SaO_2, which should be determined experimentally for each specific type of pulse-oximeter sensor: R and SaO_2 in extracted arterial blood are measured simultaneously in several healthy persons, each with several values of SaO_2. The relationship between R and SaO_2 (in the form of Equation 4) is obtained by best-fit analysis of R and SaO_2 values, measured in the calibration process. We hypothesized that the calibration process, which is based on statistical grounds, is responsible, at least partly, for the discrepancy between the pulse-oximetry output, SpO_2, and SaO_2.

Several techniques have been proposed to obviate the need for calibration. Reddy et al[58] suggested a method based on a mathematical model for the attenuation of light passing through the soft tissue, bone, and blood of a finger. Based on the model, SpO_2 is derived from the amplitudes and slopes of the PPG pulses in red and infrared and the extinction coefficients for HbO_2 and Hb. Examinations performed on healthy volunteers and patients showed agreement with a commercial pulse oximeter. Another calibration-free method based on frequency-modulated near-infrared spectroscopy (NIRS) was suggested.[14,59] However, both techniques are based on mathematical models that match tissue circulation only in approximate terms.

As explained earlier, the relationship between R and SaO_2 cannot be derived by analyzing the PPG signals in two wavelengths in red and infrared (using the Beer–Lambert law and the different absorption spectra in HbO_2 and Hb), because of the difference in light scattering between wavelengths in red and infrared. If the two wavelengths are close enough to each other so that the difference between their path-lengths can be neglected, it is possible to analytically derive the relationship between the ratio R and SaO_2:[6,9,60]

$$SaO_2 = \frac{\varepsilon_{d1} - R\varepsilon_{d2}}{R(\varepsilon_{02} - \varepsilon_{d2}) + (\varepsilon_{d1} - \varepsilon_{01})} \qquad (5)$$

where ε_o and ε_d are the extinction coefficient values for HbO_2 and Hb, respectively. The indices 1 and 2 refer to the two wavelengths. The form of Equation 5 is similar to that of Equation 4, but the coefficients of R are known: the extinction coefficient values were measured in hemolyzed extracted blood by several research groups.[61–64]

Equation 5 enables the derivation of SaO_2 from the measured parameter R and the values of the extinction coefficients with no need for calibration. This was shown by Nitzan et al,[60] using two infrared light-emitting diodes with emission spectra that peaked at wavelengths of 767 and 811 nm. The SaO_2 values, using Equation 5, were in the range of 90%–100%, while SpO_2 values obtained by commercial pulse oximeters (using red and infrared light and calibration) were 96%–98%. Higher accuracy was achieved in another study,[65] in which the light-emitting diodes were replaced by infrared laser diodes with narrow-emission spectra, and the PPG pulses were analyzed by an improved technique. The SpO_2 values measured by the two infrared wavelengths were in the range 95.3%–100.5%, and the difference between them and a commercial pulse oximeter for each examinee was 2% or less. The results of these preliminary studies provided proof of concept, but further development would be needed to make the technique clinically practical.

A similar calibration-free method based on three wavelengths in the infrared range was also suggested by the same group.[16] The use of three adjacent wavelengths obviates the need for the assumption that the difference between the path lengths of the two wavelengths can be neglected, but the method has yet to be tested and validated.

Apart from being at an early stage of development, based on analysis of the Beer–Lambert Law, calibration-free techniques are not free of flaws. A common problem in calibration-free techniques is the need for accurate values of hemoglobin extinction coefficients in order to derive SaO_2 from the PPG pulse parameters (such as R). The extinction coefficients values for HbO_2 and Hb can be found in the literature[61–64] for the wavelengths in the visible and infrared regions, but the discrepancy between the different sources is significant when aiming for accuracy of about 1%. This subject was treated by Kim and Liu[64] with respect to NIRS measurements. It should be emphasized that a lack of accurate values of hemoglobin extinction coefficients does not affect the available technique of pulse oximetry, which is based on calibration.

Motion-artifact reduction and other technical achievements

Motion artifacts can reduce the reliability of SaO_2 measurement, and are mainly important in pediatric patients and for monitoring during exercise and activities of daily living.

Several companies have developed techniques for the elimination of motion artifacts in pulse oximeters, and since the subject has been reviewed and discussed in several articles,[7,30,66,67] it is not discussed in this review. Motion rejection is generally achieved using various algorithms for differentiation between pure PPG signals and those contaminated by motion noise, but also through the introduction of improved hardware. Advances in PPG-signal analysis that are not related to pulse oximetry, such as the perfusion index and PPG variability, are also beyond the scope of the current review (see Cannesson and Talke).[68]

Venous blood oxygen saturation

Venous blood oxygen saturation (SvO_2) has physiological and clinical diagnostic significance, because a low SvO_2 value in a specific tissue combined with a normal SaO_2 value indicates reduced blood flow to that tissue, and the arteriovenous oxygen–saturation difference (SaO_2–SvO_2) is related to the balance of oxygen supply and demand in the tissue.

Similar to the measurement of SaO_2, pulse oximetry can also be used for the measurement of SvO_2, utilizing the difference in light-absorption spectra for HbO_2 and Hb. The isolation of light absorption in venous blood can be achieved by measuring the change in light absorption (in two wavelengths) following change in venous blood volume, induced either spontaneously or manually. However, in venous pulse oximetry, the scattering effect cannot be dealt with by in vitro calibration as in arterial pulse oximetry. While in vitro calibration can be performed in extracted arterial blood, because oxygen saturation has the same value in the whole arterial system, calibration by extracted venous blood cannot be applied to SvO_2 measurement, since blood extracted from a specific large vein does not necessarily have the same oxygen-saturation value as that of small veins in the tissue site, where oximetry measurement is performed.[16,60]

Some researchers have utilized respiratory blood volume changes, assuming that these changes are venous in origin. SvO_2 was derived from these changes based on previously derived empirical calibration for SaO_2. Walton et al[69] used an esophageal reflectance pulse-oximetry probe in cardiac surgery patients undergoing positive pressure ventilation, and Thiele et al[70] used a reflectance pulse-oximetry probe placed directly over three veins in volunteers. Both measured the absorbance curves of red and infrared light, and extracted blood volume changes in respiratory frequency by frequency-domain or time-domain analysis. Some algorithms yielded saturations around 80%, which is within the venous oxygen-saturation physiological range.

Another technique used for isolating the absorption effect from the combined effects of absorption and scattering of light during its pass through the tissue is NIRS, a noninvasive optical technique for the determination of the concentrations of Hb and HbO_2 and oxygen saturation in tissue. NIRS is based on the measurement of light transmission through the tissue at several wavelengths and derivation of the absorption constant at those wavelengths. The elimination of the scattering effects is done by means of several techniques, such as time-resolved spectroscopy and frequency-domain spectroscopy.[13,14] In order to derive tissue oxygen saturation by NIRS, these techniques are supported by a mathematical model, such as the semi-infinite homogeneous model. Since the matching of the model to the examined tissue that is generally heterogeneous is not perfect, the results show significant errors when applied to measurements on living tissue.[13,71–73]

In order to isolate venous blood from arterial blood, NIRS was used together with venous occlusion by a pressure cuff[74,75] or by hand.[76,77] NIRS was also combined with measurements of oscillatory blood volume changes induced by spontaneous respiration[78,79] or during mechanical ventilation,[80] assuming that the oscillatory components of blood volume changes at the breathing rate are mostly of venous origin.

Venous occlusion by a pressure cuff to increase venous blood volume was also used with measurements of light transmission in two adjacent wavelengths[60] for the assessment of SvO_2. The technique is based on the assumption of similar path-length values for the two wavelengths, and is a modification of the calibration-free pulse oximetry for the measurement of SaO_2 described in the section "Calibration".

Conclusion

Pulse oximetry has been shown to be a useful noninvasive tool for evaluation of the respiratory system since its introduction about 30 years ago. Since that time, significant technological advances in commercially available pulse oximeters have been achieved, enabling better diagnosis and monitoring of patients. The great success of pulse oximetry masks the fact that it is still burdened by an inherent potential error of 3%–4% in measurements carried out on critically ill patients and preterm newborns. It seems that the inaccuracy problem is inherent in the current technology, and significant improvement in accuracy can be achieved only through a fundamental modification of pulse oximetry. In the current review, we hypothesized that at least partly, the low level of accuracy in pulse oximetry can be attributed to the empirical calibration that is essential for the execution of conventional

pulse oximetry. It is possible that calibration-free pulse oximetry can provide SaO$_2$ measurements of higher accuracy, but there is no evidence to support this at present.

Disclosure

The authors report no conflicts of interest in this work.

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