12

Pulse Oximetry: The Working Principle, Signal Formation, and Applications

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Abstract

Pulse oximeters are routinely used in various medical-grade and consumer-grade applications. They can be used to estimate, for example, blood oxygen saturation, autonomic nervous system activity and cardiac function, blood pressure, sleep quality, and recovery through the recording of photoplethysmography signal. Medical-grade devices often record red and infra-red light-based photoplethysmography signals while smartwatches and other consumer-grade devices usually rely on a green light. At its simplest, a pulse oximeter can consist of one or two photodiodes and a photodetector attached, for example, a fingertip or earlobe. These sensors are used to record light absorption in a medium as a function of time. This time-varying absorption information is used to form a photoplethysmography signal. In this chapter, we discuss the working principles of pulse oximeters and the formation of the photoplethysmography signal. We will further discuss the advantages

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and disadvantages of pulse oximeters, which kind of applications exist in the medical field, and how pulse oximeters are utilized in daily health monitoring.

Keywords

Pulse oximetry · Photoplethysmography · Oxygen saturation · Application

Abbreviations

AC	Alternating current
AHI	Apnea-hypopnea index
C	Concentration
COHb	Carboxyhemoglobin
DC	Direct current
ECG	Electrocardiography
EEG	Electroencephalogram
H_2O	Water
HF-AC	High-frequency alternating current
HRV	Heart rate variability
ICU	Intensive care unit
IR	Infrared
I_{trans}	Transmitted intensity
LED	Light-emitting diode
LF-AC	Low-frequency alternating current
MetHb	Methemoglobin
OHb	Oxygenated hemoglobin
OSA	Obstructive sleep apnea
PPG	Photoplethysmography
PRV	Pulse rate variability
PTT	Pulse transit time
RHb	Deoxygenated hemoglobin
SaO_2	Arterial oxygen saturation
SpO_2	Peripheral blood oxygen saturation

12.1 Working Principle

Pulse oximetry is a method initially developed for the measurement of peripheral blood oxygen saturation (SpO_2). It is an optical technique based on differences in light absorption spectra of oxygenated (OHb) and deoxygenated (RHb) hemoglobin (Nitzan et al., 2014). More precisely, the estimation of the SpO_2 is based on photoplethysmography (PPG, see Sect. 12.2). As a noninvasive method, having a high correlation with invasive arterial oxygen saturation (SaO_2) defined based on arterial blood gas analysis (Nitzan et al., 2014), it has become a valuable technique for measuring oxygen saturation in clinical settings.

In addition to being noninvasive, pulse oximetry has several other advantages. It is a safe, comfortable, and inexpensive method with no need for end-user calibration. As the oximeter is usually placed to the fingertip or earlobe in a medical setting, it can usually be self-applied and does not require a medical specialist. Furthermore, various consumer-grade health technology solutions, such as smartwatches and smartphones, are also capable of estimating SpO_2 with a reasonable correlation to SaO_2. As pulse oximetry is a simple and inexpensive method already integrated into various settings, it is ideal for long-term monitoring of overall well-being, stress, recovery, quality of sleep, and based on the recent research, also for detection of sleep disturbances and disorders.

Pulse oximetry can rely either on the transmission or the reflection of light (Fig. 12.1). Out of these, transmissive pulse oximetry is the most common in medical devices. In transmissive pulse oximetry, light sensors are placed usually on a fingertip or earlobe. The response time of conventional oximeter probes varies and, for example, ear probes respond quicker to a change in blood oxygen saturation than finger probes (Young et al., 1992). Also, the sensors can be attached to a toe in newborns. The sensors emit light usually with two different wavelengths, and the light passes through the skin and reaches a photodetector that measures the changes in absorption of both wavelengths. The reflective mode, on the other hand, can be applied to different parts of the body, not only on the fingertips or the thin portion of the ear, to measure the saturation and PPG signal. Similarly, as in the transmissive pulse oximetry, the sensors emit light with two different wavelengths. However, the main difference is that in reflectance pulse oximetry the photodetector is located next to the light-emitting sensors and detects reflected and backscattered photons of both emitted wave-

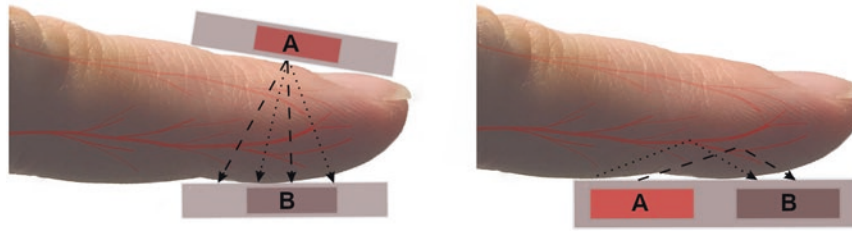

Fig. 12.1 Obtaining the photoplethysmogram and saturation signal with a pulse oximeter. A transmissive mode, on the left-hand side, and a reflective mode, on the right-hand side, are presented. A: a light source (emitting, for example, red and infrared light) and B: a detector

lengths. In some cases, pulse oximeters may utilize more than two different wavelengths of light.

Pulse oximetry is widely used in different clinical domains and is, for example, a basic measurement at intensive care units (ICUs) and always included in polysomnographic evaluations. Pulse oximetry is mainly used to measure SpO_2, but the measured PPG signal contains a vast amount of information and works as a proxy for several physiological functions. For example, it can be used to evaluate indirectly blood pressure (Nachman et al., 2020), depth of anesthesia (Shelley, 2007), pulse rate (Shelley, 2007), heart rate variability (Gil et al., 2010), cardiac arrhythmias (Blanc et al., 1993), respiratory rate (Shelley, 2007), sleep stages (Huttunen et al., 2021), presence of sleep disorders (Lazazzera et al., 2021; Nikkonen et al., 2019), and sleep disorder-related daytime symptoms (Kainulainen et al., 2020a, b). The sampling frequency of a pulse oximeter depends on its intended use and is typically between 1 and 256 Hz. For the determination of SpO_2 in clinical settings, 10–25 Hz sampling frequency is currently recommended (Iber et al., 2007). However, a minimum sampling frequency of 200 Hz is required to get more accurate estimation on, for example, pulse rate and heart rate variability from PPG signal (Béres et al., 2019). Thus, the typical sampling frequency for PPG signal acquisition is 256 Hz.

12.1.1 Green, Red, and Infrared Light

Transmissive pulse oximeters with two light sources are the most commonly used type of oximeters in clinical practice and their working principle is based on the light absorption in the peripheral arterial blood (Fig. 12.2), more specifically, the absorption induced by the OHb and RHb (Damianou, 1995; Mannheimer, 2007). When oxygen binds with the iron ion in the blood, the structure of the heme group in the binding site changes from a non-planar to a planar orientation (Benesch et al., 1975). These molecular structure changes with oxygenation result in differences in light absorption between OHb and RHb (Fig. 12.2).

As illustrated in Fig. 12.2., at 660 nm (red light), the absorption of light is mainly caused by RHb in blood. In contrast, at 530 nm and 900 nm (green and infrared, respectively), the absorption of light is mainly caused by OHb as the molar extinction coefficient of OHb is higher compared to that of RHb. The green, red, and infrared light can penetrate through soft tissue (mostly consisting of water) and the amount of transmitted light can be measured. This information can be used to estimate the level of oxygen in the blood. The measurement is, of course, disturbed by the light scattering within the medium and reflections from the light source-skin surfaces, but these interactions are not significant (Tuchin, 2015).

In transmissive pulse oximeters, red light is more often used than green light. This is because

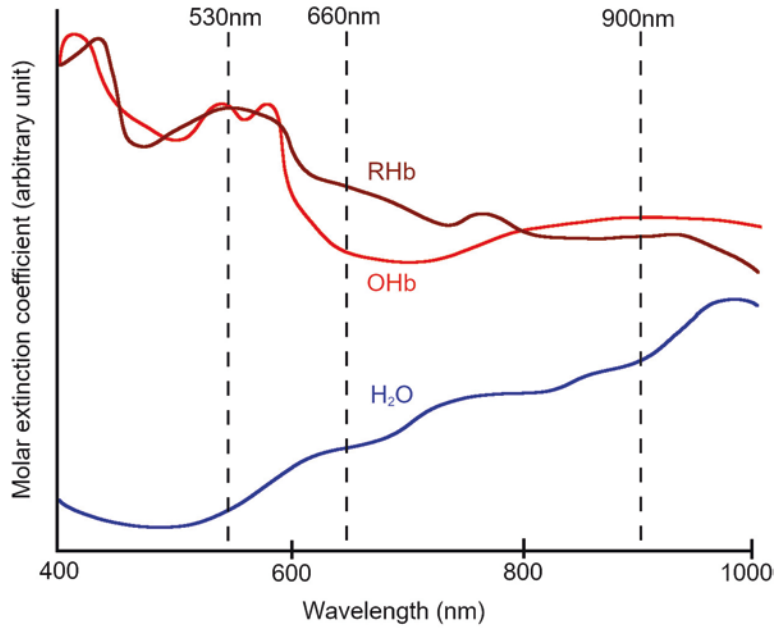

Fig. 12.2 Illustration of the logarithmic molar extinction coefficients as a function of light wavelength. 530 nm, 660 nm, and 900 nm correspond to the green, red, and infrared light wavelengths commonly used in pulse oximeters, respectively (Tuchin, 2015; Kainulainen, 2020). OHb oxygenated hemoglobin, RHb deoxygenated hemoglobin, H_2O water

the use of red light increases the accuracy and precision of the measurements of, for example, heart rate and blood oxygen saturation. Also, the human body poorly absorbs red light allowing it to penetrate much deeper than green light. Therefore, the usefulness of green light in determining muscle saturation or total hemoglobin is limited. Also, red light is not affected as much by dark skin tone or tattoos which can distort the measurements done with green light. However, consumer-grade devices are most often reflective pulse oximeters utilizing green light to measure PPG. This is because the use of green light has several benefits. First, green PPG amplitudes are the strongest across the range of visible light (Verkruysse et al., 2008). Second, the human tissues are good absorbers of green light and, thus, green light coming from external sources does not disturb the measurement affecting the signal quality.

Pulse oximeters utilizing more than two wavelengths (usually four) also exist. They are usually transmissive pulse oximeters and can differentiate between more than two blood components. This is, they can detect other hemoglobin types in addition to RHb and OHb, such as carboxyhemoglobin (COHb) and methemoglobin (MetHb) also called dyshemoglobins. Therefore, multiple wavelength pulse oximeters are often called CO-oximeters (Zaouter & Zavorsky, 2012). Four-wavelength CO-oximeters are typically used as a gold standard reference when evaluating the accuracy of standard oximeters as they are more accurate (Sinex, 1999) or when the standard oximeter is not sufficient, for example, when carbon monoxide poisoning is suspected.

12.2 Photoplethysmogram

With transmissive pulse oximeters, an absorption signal is formed at a photodetector which measures transmitted light passed through a medium. For example, soft tissue, bony structures, and blood absorb light differently and these structures affect the absorption signal characteristics when measuring the absorption signal from a fingertip. Cardiac and respiratory functions cause changes in the blood volume and composition in the arterial blood. Therefore, the concentrations of OHb and RHb, and the optical path length of the light change constantly as a function of time (Damianou, 1995; Mannheimer, 2007). This light absorption signal varying as a function of time is called a photoplethysmogram.

The wavelength of the light affects the PPG signal, i.e., the absorbance. Red light is more sensitive to changes in oxygenation than infrared light as the molar extinction coefficient of RHb is higher and water absorption lower (Fig. 12.2). The absorption of infrared light is similar to water and both oxygenated and deoxygenated hemoglobin. Moreover, the total absorption is lower with infrared light compared to red light. Therefore, the PPG signal measured with infrared wavelength is more stable and is more commonly used than red light (Alian & Shelley, 2014). However, both red and infrared light measurements are needed to estimate SpO_2.

The absorbance signal is formed by three components: the direct current (DC), the low-frequency alternating current (LF-AC), and the high-frequency alternating current (HF-AC) components (Fig. 12.3). The DC component represents the absorption of light in static mediums and has no pulsatile time-varying component. In contrast, the LF-AC and HF-AC vary temporarily. The LF-AC component carries information on changes in blood volume which can be caused by alterations in breathing, thermal regulation, autonomic nervous system activity, and hemoglobin concentration. The HF-AC component mainly represents the pulsatility of the arteries and forms the basis of the PPG signal.

One PPG waveform can be separated into systolic and diastolic phases (Gamrah et al., 2020). The systolic phase starts when the aortic valve opens. During the systolic phase, blood pressure increases as blood flows into the aorta. This can be seen as a decrease in the transmitted light intensity as a function of time. After the systolic maximum pressure is reached (the first maximum peak), the blood pressure starts to decrease and the aortic valve closes. When the aortic valve closes, another increase in the blood pressure can be seen (the second maximum peak) (Hogan et al., 2014). Between these two maximum peaks is the Dicrotic Notch representing the end of the systolic phase and the start of the diastolic phase (Gamrah et al., 2020). During the diastolic phase, blood flows out from the aorta and blood pressure decreases (transmitted light intensity increases as a function of time) until the minimum diastolic pressure is reached. Then, the aortic valve opens again, and a new cycle starts from the beginning.

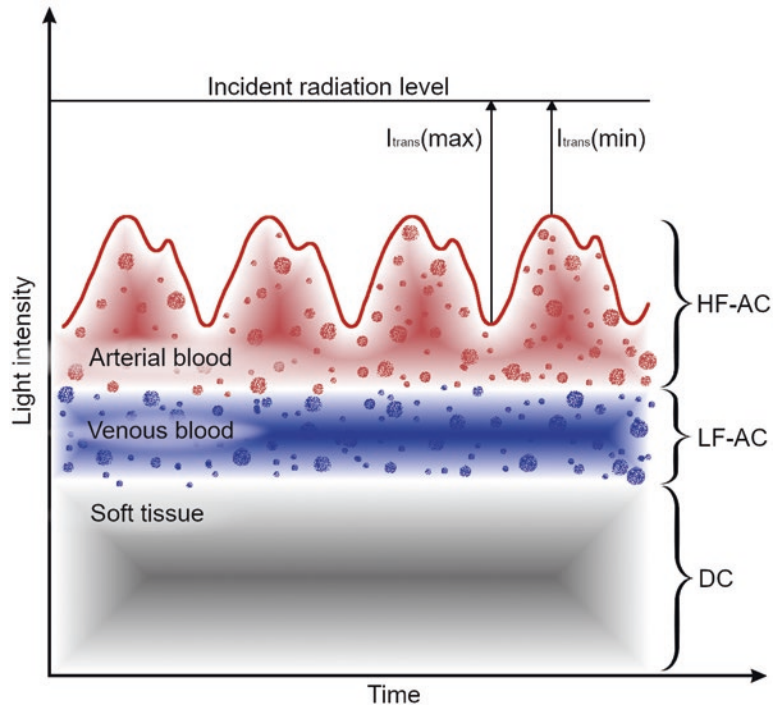

Fig. 12.3 Illustrative example of the formation and waveform of photoplethysmography signal. (Figure modified from Tiihonen (2008), Kainulainen (2020). AC alternating current, DC direct current, LF-AC low-frequency AC component, HF-AC high-frequency AC component, I_{trans}(max) transmitted intensity when the optical path length is the longest, I_{trans}(min) transmitted intensity when the optical path length is shortest

However, it has to be noted that the strengths of both LF-AC and HF-AC components are only a few percentages from the total absorption.

Green light with a wavelength of 530 nm cannot penetrate as deep in tissue as red (660 nm) or infrared light (900 nm) (Huang et al., 2011). The green light can only penetrate through the epidermis and papillary layer while red and infrared lights can reach the dermis, subcutaneous, and deeper layers (Sviridova et al., 2018). When using red or infrared light, the PPG waveform is mainly determined by blood volume changes in the peripheral arteries. However, especially with large vessels, green light cannot penetrate deep enough to reach the arteries, and information on blood volume and flow changes cannot be directly obtained. Therefore, contrary to what has been thought before (Damianou, 1995; Mannheimer, 2007), it has been proposed that the PPG signal is formed based on the pulsatile transmural pressure of the arteries, i.e., the pressure difference between walls of the thinnest capillaries in the papillary dermis (Kamshilin et al., 2015). The distance between nearby capillaries changes as the intercapillary tissue stretches and compresses caused by the beating of the heart. Thus, changes in the density of the capillaries in the papillary dermis lead to changes in the optical properties of the tissue (Kamshilin et al., 2015; Volkov et al., 2017).

12.2.1 Blood Oxygen Saturation

The blood oxygen saturation is defined as a ratio between concentrations of OHb and RHb:

$$\mathrm{SpO}_2 = \frac{C(\mathrm{OHb})}{C(\mathrm{OHb}) + C(\mathrm{RHb})} \times 100\%$$

where C is the concentration. However, invasive measurement is required for accurate definition of the concentrations and, thus, in noninvasive transmissive mode applications, the estimation of SpO_2 is based on the absorption of red and infrared light. PPG signals of red light and infrared light oscillate at the same phase, but the PPG signal of infrared light has a smaller amplitude. During one heart cycle, hemoglobin concentration and body temperature are nearly constant; thus, the only time-varying component is the optical path length of light that is mainly affected by the arterial blood volume changes. This leads to a situation where the AC component (Fig. 12.3), for both PPG signals, can be defined based on the differential absorption during one pulse wave (Damianou, 1995; Mannheimer, 2007). By further dividing the differential absorption of red light with differential absorption of infrared light, the SpO_2 value can be estimated (Damianou, 1995, Mannheimer, 2007). To summarize, both red and infrared PPG signals are required to estimate peripheral blood oxygen saturation noninvasively.

12.2.2 Pulse

Although pulse oximeters are colloquially said to also measure heart rate, this is not strictly true. Instead, pulse oximeters can only record pulse rate (Schäfer & Vagedes, 2013). Generally, the pulse rate and heart rate are highly correlated. Thus, in most everyday applications, heart rate can accurately be estimated by pulse rate. However, minor differences do exist. Since the pressure wave generated by the heart will take a short time to reach the fingertip, there is a small delay between the systole and the time the pulse is detected at the periphery (Smith et al., 1999). This time delay is also called the pulse transit time (PTT), and it can be accurately recorded with simultaneous electrocardiography (ECG) and PPG recording (Smith et al., 1999). Since the pulse through the artery is a mechanical wave, it travels at the speed of sound, making the PTT very short, but still significant. PTT varies between study subjects and is affected by the elasticity of the artery wall and blood pressure among many other factors (Mukkamala et al., 2015). Thus, PTT can also be used for various purposes such as continuous blood pressure estimation (Mukkamala et al., 2015; Smith et al., 1999).

Heart rate variability (HRV) analysis is a common method for evaluating autonomic nervous system function (Pinheiro et al., 2016; Hietakoste et al., 2020). HRV parameters are obtained from

ECG by detecting the fiducial point of R waves and generating the R-R intervals. Since ECG signal is not always available during sleep recordings and there can be artifacts influencing R wave detection, the information on HRV can be also obtained by using the pulse signal. As the detection of the exact time point of the R-peak is important, accurate pulse peak detection is equally essential. However, as the pulse wave is relatively wide (Fig. 12.3) and not as sharp as the R-peak, there is intrinsically more inaccuracy in the pulse wave detection (Schäfer & Vagedes, 2013). However, since there are factors beyond cardiac electrophysiology affecting the PPG signal (Gil et al., 2010), pulse rate variability (PRV) is not directly comparable to HRV. Still, it can be used as an indicator to quantify the activity of the autonomic nervous system (Pinheiro et al., 2016; Mejía-Mejía et al., 2020).

12.3 Error Sources and Limitations

Even though pulse oximetry is a powerful tool, it also has certain limitations (Table 12.1) some of which are discussed in this section. One major limitation of pulse oximetry is that it measures the ratio of oxygenated and deoxygenated hemoglobin rather than ventilation or the amount of oxygen in the tissue (Mcmorrow & Mythen, 2006). While these are normally tightly linked in a healthy subject, some conditions can cause hypoxia in the tissue level even if the measured

Table 12.1 Possible error sources of a pulse oximeter

Error source	Type of error	Reference
Poor peripheral circulation Hypotension Hypothermia/cold periphery Vasoconstriction	Intermittent drop-outs or inability to read SpO_2 or false readings	Hakemi and Bender (2005), Chan and Chan (2013)
Movement artifacts Especially when estimated from a fingertip	Falsely low SpO_2	Mcmorrow and Mythen (2006), Tsien and Fackler (1997), Chan and Chan (2013)
Venous pulsation Probe too tight around the finger Severe tricuspid regurgitation Heart failure	Falsely low SpO_2	Chan and Chan (2013)
Dyshemoglobinemia Carboxyhaemoglobinaemia Methemoglobinemia	False readings	Tremper (1989)
Carbon monoxide poisoning	Falsely normal or elevated SpO_2	Vegfors and Lennmarken (1991), Barker and Tremper (1987)
Lightning Light-emitting diodes, infrared, ultraviolet, fluorescent lamps (the effect can be tested by covering the probe)	False readings	Amar et al. (1989), Schulz and Ham (2019)
Light barriers Nail polish, artificial nails (influence depends on the color) Tattoos Skin discoloration (caused by tobacco, dirt, or paint)	False readings	Coté et al. (1988), Samman et al. (2006), Ralston et al. (1991b)
Dark skin tone	Possible false readings	Ries et al. (1989, Cecil et al. (1988), Ralston et al. (1991b), Bickler et al. (2005)
Severe anemia	Falsely low SpO_2 in hypoxemic patients	Chan and Chan (2013)

SpO_2: peripheral blood oxygen saturation

oxygen saturation is normal. For example, undetected major loss of blood (e.g., internal bleeding) may cause a lack of oxygen in the tissue level, even though the SpO_2 reading is still high. Severe anemia can cause a similar condition. However, it is critical to note that in these cases, the pulse oximeter reading is not false and the blood can be highly oxygenated; there may just not be enough blood volume to keep the tissue fully oxygenated (Mcmorrow & Mythen, 2006).

Pulse oximetry does not require calibration before use for different individuals (Ralston et al., 1991a). The zero calibration is performed automatically due to measuring light absorption both during the diastolic and systolic phases. Gain calibration is not required either as the absolute light intensity at the photoreceptor does not affect the estimated oxygen saturation. Only the ratio of the absorbed light in the two (or more) wavelengths is considered. While the lack of calibration is certainly one of the greatest advantages of pulse oximetry, it is also a major error source. As no calibration is needed, the device simply uses a pre-determined look-up table to estimate the peripheral oxygen saturation based on the ratio of the absorbed light (Ralston et al., 1991a). The values in the look-up table are derived experimentally by simultaneous measurement of true SaO_2 and the light absorbance (Jubran, 1998). An absorption ratio curve can then be formed based on these measurements (Sinex, 1999). Thus, the result of using this standardized look-up table is that if some unaccounted factors are affecting the light absorption ratio other than the fraction of oxygenated hemoglobin, the pulse oximeter can give false readings. In addition, the population in which this curve is produced, i.e., the calibration population, will affect the ultimate accuracy of the pulse oximeter (Sinex, 1999; Mcmorrow & Mythen, 2006). As the experimental data cannot be obtained for arbitrarily low saturation values, the accuracy of pulse oximeters decreases with lower saturation (Sinex, 1999; Jubran & Tobin, 1990). Accuracy rates reported for individual instruments are often in the range of ±2–3% (Balady et al., 2010). Wider confidence limits are not unusual, particularly in the saturation less than 85%, but SpO_2 values of less than 70% are usually considered unreliable (Mcmorrow & Mythen, 2006).

The theoretical foundation behind transmissive pulse oximetry assumes that the blood contains only oxygenated and deoxygenated hemoglobin, which is strictly not true as the hemoglobin can also form other compounds such as carboxyhemoglobin or methemoglobin (Ralston et al., 1991a; Zaouter & Zavorsky, 2012). These compounds, also called dyshemoglobins, affect the absorption and, thus, affect the oximeter reading. Normally, the dyshemoglobins have little effect on the oximeter reading, but in some cases, they can be a major source of error (Tremper, 1989). If the person is suffering from carbon monoxide poisoning, the amount of carboxyhemoglobin in the blood is high. However, standard pulse oximeters cannot reliably distinguish between carboxyhemoglobin and oxygenated hemoglobin (Vegfors & Lennmarken, 1991). True oxygenation of as low as 30% can still give >90% saturation readings when carboxyhemoglobin concentration in the blood is high (Barker & Tremper, 1987). Smoking a cigarette also increases the level of carboxyhemoglobin and can thus temporarily elevate the pulse oximeter reading (DeMeulenaere, 2007). If the concentration of methemoglobin is high, it can dominate the absorption spectrum and erroneously cause a reading of around 85% regardless of the true saturation (Barker et al., 1989). The errors caused by the presence of dyshemoglobin in blood could be eliminated by using multi-wavelength CO-oximeters. However, due to their higher cost, they are commonly only used in specialized applications, such as when carbon monoxide poisoning is suspected or as a reference (Sinex, 1999).

Newborn infants have a different type of hemoglobin in their blood called fetal hemoglobin. It has a different composition and bonds to oxygen more strongly than adult hemoglobin. However, it has a similar absorption spectrum in the wavelength 650-1000 nm (Ralston et al., 1991b). As this is the wavelength range most two-wavelength oximeters operate in, the presence of fetal hemoglobin has little effect on the oximeter reading. However, the fetal hemoglobin

will affect the reading of a multi-wavelength CO-oximeter and, thus, its presence should be considered and the oximeter reading corrected (Ralston et al., 1991b).

As the transmissive pulse oximeter assumes that the other tissue and venous blood absorption are constant, limb motion can also cause movement artifacts and is a significant error source (Mcmorrow & Mythen, 2006). Movement artifacts are the most common type of error in the oximeter reading and have been reported to cause as much as half of the false alarms in pulse oximeters in the ICUs (Tsien & Fackler, 1997). In a diagnostic test setting (for example during a cardiac-pulmonary test), the amount of movement artifacts can be reduced by placing a sensor on to forehead skin. Furthermore, excess sweating can affect SpO_2 readings; however, the effect of sweating is easily reduced by wiping the electrode.

Pulse oximetry can also be affected by skin tone and the accuracy of measurements can be decreased in persons with dark skin tone (Ries et al., 1989; Cecil et al., 1988; Ralston et al., 1991b), although contradicting results have also been reported (Bothma et al., 1996; Adler et al., 1998). However, a slight decrease in accuracy due to skin tone is not usually clinically significant (Cecil et al., 1988). Tattoos, artificial nails, and nail polish can also decrease the pulse oximeter accuracy as they can change the light absorption (Jubran, 1999; Coté et al., 1988; Ralston et al., 1991b). However, these issues are usually easily avoided simply by using a non-tattooed skin surface and by removing nail polish or artificial nails.

Low perfusion can also cause errors in pulse oximetry readings (Hakemi & Bender, 2005). Low perfusion could be caused by vasoconstriction, low cardiac output, or even hypothermia. Extremities can be warmed with a mitten (for example during tilt table test) to maintain a reliable transmittance of pulse oximetry and PPG signals. In a case of low perfusion, it might be difficult for the pulse oximeter to reliably detect the HF-AC component of the signal from the noise and, thus, may give erroneous readings. To be considered reliable, the pulse signal from the oximeter should be regular and match the heart rate from the ECG if available. Similarly, if a subject has very low blood pressure, the pulse oximeter readings should be considered unreliable (DeMeulenaere, 2007).

The components used in the pulse oximeters may affect the accuracy of the device, especially if the components are not of the highest quality (Mcmorrow & Mythen, 2006). For example, cheap light-emitting diodes (LEDs) may emit a slightly different frequency of light as designed or the photodetectors may report slightly erroneous intensities. These error sources could be eliminated by only using high-quality components but should still be accounted for especially in consumer-grade devices.

Although there are several limitations in pulse oximetry, the error sources can usually be accounted for or corrected if the error sources are acknowledged. However, a considerable proportion of users, either healthcare professionals or consumers, may not be aware of what exactly a pulse oximeter is recording and that the pulse oximeter is not a direct measure of partial pressure of oxygen in the blood (Stoneham et al., 1994). Therefore, some of the limitations in pulse oximetry may be magnified by these misconceptions as the error sources may not be acknowledged and, thus, false readings may not be noticed.

12.4 Applications

12.4.1 Consumer Use

Consumer pulse oximeters are widely available. Most of these low-cost pulse oximeters have not been rigorously tested and validated against clinical measurement and do not meet standards for medical devices. However, most devices can safely rule out hypoxemia in the vast majority of patients (Harskamp et al., 2021). This is important since the demand for oximeters has increased during the COVID-19 pandemic, and the World Health Organization has recommended home oximetry monitoring for patients with COVID-19 and with risk factors for progression to severe

disease (Organization, 2021). Simultaneously, PPG recording has gained popularity in consumer-grade wearable devices. The solutions come with multiple forms and arrangements depending on the manufacturer, but most often the basic principles are the same. Wearables most often utilize reflective mode and green light, or a combination of different light colors. Vulcan et al. (2021) published an extensive review that summarizes the current status of wrist-worn consumer-grade products for sleep tracking. The study reveals that devices perform well when considering the basic PRV estimation, heart rate tracking, and other physical activity, especially if the tracker is equipped with an accelerometer. When tested against PSG measurements, most of the consumer-grade sleep trackers perform relatively poor in epoch-by-epoch sleep staging (Vulcan et al., 2021; Chinoy et al., 2021). Furthermore, it has to be noted that most of the sleep trackers use multiple information sources for the sleep stage estimation, not just the PPG. However, they can measure total sleep time and differentiate wake from sleep with sufficient accuracy surpassing, for example, plain actigraphy. Therefore, it could be speculated that these devices would fit well in long-term evaluations together with sleep diaries.

In the clinical setting, the PPG is most often measured with transmissive mode from the fingertip or earlobe. In contrast, most of the trackers are wrist-worn devices and exploit reflective mode. A variety of studies show that the measurement location affects the measured signal quality and the achieved correspondence with the gold standard measurement. For example, the respiratory rate, heart rate, and RR-interval evaluation are not as accurate from the wrist with reflective mode oximeters as from the fingertip with transmissive mode oximeters (Hartmann et al., 2019; Longmore et al., 2019).

The PPG-based consumer-grade approaches show a lot of potential but most of them lack validation against the gold standard measurements. Other disadvantages are the non-harmonized measurement techniques, measurement locations, and signal processing algorithms (Vulcan et al., 2021). However, it has to be noted that these devices are designed for commercial, not medical use, and, therefore, the performance demands are very different.

12.4.2 Clinical Use

A pulse oximeter is a vital part of standard clinical assessment. It is routinely used in emergency rooms and patient wards because it can quickly give information on ventilation and perfusion deficits. Saturation measurement is also an important tool for diagnostics. For example, pulse oximeter measurements are obtained routinely during cardiopulmonary exercise testing (Balady et al., 2010). A decrease in saturation by >5% is commonly used as an indication of pulmonary limitation to exercise (Balady et al., 2010). Normally, the SpO_2 of a healthy subject is at least 95%, while values <90% are considered low and possibly alarming (Broaddus, 2016; Duncan, 2017). The most common causes for decreased SpO_2 levels are ventilation-perfusion mismatch (for example due to apnea or obstructive pulmonary disease), cardiovascular shunt, abnormal pulmonary diffusion capacity, hypoventilation, or decreased oxygen content of the inhaled air (Bhutta et al., 2021).

The SpO_2 is constantly measured and followed in the ICU. However, critically ill patients' pulse oximeter-based oxygenation estimation has been criticized for its tendency to overestimate the readings (Van de Louw et al., 2001). Also in these patients, periphery vasocontraction is prevalent. For example, low perfusion and sepsis among other conditions can lead to over 90% pulse oximeter readings, even though the true SpO_2 is below 90% (Wilson et al., 2010). For these reasons, ICU perfusion is often screened with more robust methods. Global oxygenation may be also monitored intermittently through blood gas analysis or continuously with specialized catheters (Kipnis & Valle, 2016).

In sleep medicine, especially when assessing obstructive sleep apnea (OSA), a pulse oximeter is a cornerstone of the measurements. When considering the SpO_2, conventional parameters such as the apnea-hypopnea index (AHI), oxygen

desaturation index, time spent under 90% saturation, mean SpO_2, and nadir SpO_2 can be estimated. Based on the latest research, the AHI can be evaluated utilizing only a pulse oximeter recording (Nikkonen et al., 2019; Leino et al., 2021; Álvarez et al., 2017). In addition, the frequency domain information in PPG and the desaturation areas can be exploited independently to assess the severity of sleep apnea beyond the AHI (Kainulainen et al., 2020a, b; Kainulainen et al., 2019; Azarbarzin et al., 2019). Thus, it could be speculated that OSA screening in various patient populations and measurement conditions could be conducted based even solely on a pulse oximeter.

However, a pulse oximeter provides a lot more information besides the SpO_2. The PPG signals contain information on the pulse rate, PRV, peripheral vasoconstriction, and perfusion (Jubran, 2015; Lima et al., 2002; Pinheiro et al., 2016; Rauh et al., 2004; Tusman et al., 2019). The recurrent breathing cessations and flow limitations cause the cardiorespiratory system to respond to changes in intrathoracic pressure and deoxygenation. The responses are further seen in the PPG signal as deviations from normal waveform, amplitude drops, and frequency modulations. For simplistic analysis and to gain a better overview of the condition of the patient, normal PRV metrics can be assessed while keeping in mind its limitations (see Sect. 12.2.2).

Moreover, the parasympathetic nervous system controls not just the cardiorespiratory system and various body functions but also sleep stages (Fink et al., 2018). For these reasons, the PPG signal carries information also on the depth of sleep. PPG could be used as a surrogate for electroencephalogram (EEG) in detecting rapid eye movement (REM) sleep, non-REM sleep, and wakefulness (Huttunen et al., 2021; Korkalainen et al., 2020). In addition to sleep stage scoring, PPG could be used for arousal detection (Karmakar et al., 2014). Arousals cause rapid changes in pulse and short periods of vasoconstriction. These phenomena are visible in the PPG signal as frequency increases and amplitude decreases, respectively. Simultaneously, the amplitude drops in PPG are surrogates for subcortical brain activity, even though EEG-based arousal is undetectable (Delessert et al., 2010). In summary, the utility of the pulse oximeter in sleep medicine is evident.

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