Improved Single-LED Pulse Oximeter Design Based on Multi-Wavelength Analysis

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***Abstract*—Being often considered as one of the most important vital signs, oxygen saturation is a key element in the management of patient care. Pulse oximetry is one of the clinical tools used to measure oxygen saturation of the arterial blood almost instantly. Nevertheless, healthcare workers know that there are some technological limitations that can impact the accuracy of these devices. Using more than two wavelengths for estimating the oxygenation can improve the accuracy but at the expense of a higher price, which can limit the use in home settings. This paper proposes a different approach for a more affordable pulse oximeter, based on spectral identification in the red and near-infrared light domains. Thus, compared to a classical pulse oximeter solution, the proposed design is able to detect abnormal levels of hemoglobin derivatives by using six different wavelengths, offering this way improved accuracy in critical situations and helping in-home patients to deal with the first signs of sickness.**

Keywords—health management; multi-wavelength sensor; oxigen saturation; pulse oximetry; pulse rate.

I. INTRODUCTION

Pulse oximeters have become a critical diagnostic tool, due to their role in a multitude of clinical scenarios [1]-[3], blood oxygen saturation level estimation being considered the fifth vital sign [4]. Pulse oximetry is a technique that determines the oxygen saturation (SpO_2) and the pulse rate (PR) in a non-invasively manner. This estimation is based on a photoplethysmogram (PPG), obtained with the help of two wavelengths of light, usually red and infrared, as these wavelengths have a better propagation through tissues [5].

Starting with the COVID-19 pandemic, the issue of blood oxygen levels monitoring became a main preoccupation, which has led to an increased interest in the use of general-purpose pulse oximeters. Such devices provide an acceptable accuracy, enabling real-time patient's condition monitoring. Wide access to reliable pulse oximeters is limited by the cost of these devices, leading to significant public health implications. As a response, low-priced instruments for pulse oximetry became available, providing a solution for individual health monitoring but also for other applications [6], [7].

In order to estimate the accuracy, the precision, and the bias of these devices, several studies have been performed on low-cost pulse oximeters, and the conclusion drawn was that most of them do not meet the medical standards required to be considered high-performance devices. Moreover, it is known [8], [9] that there are some cases in which the indications provided by such instruments may have limitations even in the professional line of devices, due to various factors. Thus, in 2021, the Food and Drug Administration (FDA) issued a safety warning that pulse oximeters have a risk of inaccuracy under certain circumstances [10], such as when abnormal levels of hemoglobin variants are present, especially when using over-the-counter (OTC) devices. As such, the need for a wide-accessible non-invasive measuring instrument for a better SpO_2 estimation is more necessary than ever.

In the upper-mentioned context, this paper presents a new pulse oximeter design aimed to provide improved accuracy and precision. Unlike most of the existing designs, the one proposed in this work is based on a measurement system with six wavelengths, which can offer a better view on the levels of abnormal hemoglobin that can impact the accuracy of SpO_2 estimation.

II. STATE-OF-THE-ART IN PULSE OXIMETRY

The oxygen saturation is an indicator of the oxygen carried in the circulatory system, being defined as the ratio between concentration of oxy-hemoglobin (HbO_2) and the total concentration of both the reduced hemoglobin (Hb) and oxy-hemoglobin, present in the blood [11]:

$$SpO_2[\%] = \frac{C_{HbO_2}}{C_{HbO_2} + C_{Hb}} \cdot 100 . \quad (1)$$

When the light passes through the subcutaneous tissues, the absorbance A is defined by:

$$A = log_{10}\left(\frac{I_0}{I_t}\right), \quad (2)$$

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where I_0 is the intensity of the incident light, and I_t is the intensity of the transmitted light.

The common expression of the Beer-Lambert Law states that the absorbance is dependent on the molar absorption coefficient ε at a specific wavelength, the molar concentration C, and the distance d traveled by light through the tissue:

$$A = \varepsilon \cdot C \cdot d \ . \tag{3}$$

The (2) and (3) show that the exponential decreasing of the intensity of light depends on distance traveled through the tissues and the wavelength of light:

$$I_t = I_0 \cdot 10^{-\varepsilon \cdot C \cdot d} \ . \tag{4}$$

The basic principles of operations for monitoring the PR and the SpO_2 are based on two wavelengths of light, λ_1 and λ_2, either in a system that relies on reflected light over the cutaneous tissue or in a system that relies on light transmission through blood-perfused tissues.

Based on the Fig. 1 [9] for absorption coefficient, the two equations that describe the absorption of light at λ_1 and λ_2 by both Hb and HbO_2 can be solved to find the relation that determines the concentration of HbO_2 and Hb in the circulatory system:

$$SO_2 = a - b \cdot \frac{A(\lambda_1)}{A(\lambda_2)}, \tag{5}$$

where a and b are functions of absorption coefficient. Equation (5) is used by the majority of pulse oximeter's makers to determine the SpO_2. These two wavelengths are chosen in zones of spectrum where HbO_2 and Hb absorb differently.

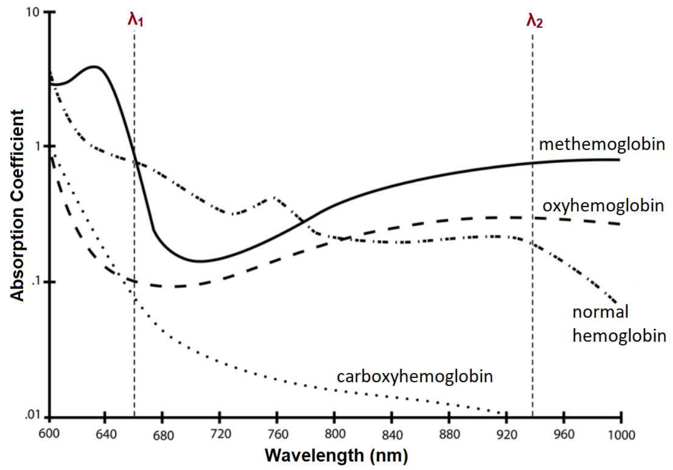

Fig. 1. Light absorbance (absorption coefficient) versus wavelength for four hemoglobin species: oxyhemoglobin (HbO_2), normal hemoglobin (Hb), methaemoglobin (HbMet), and carboxyhemoglobin (HbCO).

A. Devices Based on Reflected Light Over the Tissue

The technique of reflectance-based pulse oximetry is used for monitoring the pulse rate and the oxygen saturation on areas of the body where measurements through transmittance-based pulse oximetry cannot be carried out, such as forehead, chest or wrist. The incident light from such a device is passing through the skin and it is reflected off the tissues and bones. The biggest issue for this system is the lower signal-to-noise ratio, which makes it more challenging for the algorithm to interpret the results. However, a prototype build in [12] demonstrates that this SpO_2 measurement method can be implemented with a precision of 1%. In [13], a pulse oximeter concept based on a reflective PPG sensor with Fresnel lens was demonstrated, enabling a reflected signal enhancement higher than 140%. This can lead to a low-cost device that is very helpful in monitoring the health status without hindering the patient's day-to-day activities.

B. Devices Based on Light Transmission Through the Human Tissue

The more traditional path is the system that relies on light transmission mode, even if this method has the disadvantage that it can only be used on few body parts, such as a finger or an ear lobe. Since arterial flow is pulsatile in nature, the constant component of light absorption due to blood vessels, tissues, and bones must be subtracted, leaving only the variable values. Next, the arterial oxygen saturation is estimated using an internal algorithm created empirically in desaturation studies on healthy volunteers. The technology used by this type of pulse oximeters is based on the measurement of the absorption produced by the passage of two light rays of different wavelengths (e.g. 660 nm, which is red, and 940 nm, which is near-infrared) through the human tissue (e.g. through the finger), in direct relation with the presence of HbO_2 and Hb, and their own absorption coefficient. However, there are certain situations when these devices cannot offer accurate results, due to the intrinsic technology limitations. Indeed, as it can be seen in Fig. 1, other variations of hemoglobin, such as HbMet or HbCO, can still impact the final result when some concentration limits are exceeded. Obviously, with only two wavelengths, classic pulse oximeters cannot determine the presence of HbMet or HbCO, so different solutions have been proposed, each with their own advantages and drawbacks. For instance, the authors of [14] conducted a study based on a Masimo Rad-57, a pulse oximeter that utilizes a sensor with eight monochromatic LEDs for accurate determination of SpMet and SpCO, but still using two wavelengths for HbO_2 estimation. In [15] a pulse oximeter schematic based on a single monochrome LED and a buried quad junction (BQJ) multispectral photodiode is proposed. This solution offers an oxygen saturation estimation with a mean error of 1.3%. In [5], two IR wavelengths having similar scattering constants and path-lengths have been used in order to prevent errors in SpO_2 estimation.

III. Description of the Proposed Device

Many low-cost pulse oximeters marketed nowadays demonstrate highly inaccurate readings, with an average root mean square (ARMS) error greater than 3% for SpO_2 values [8], especially in deeper hypoxia. As demonstrated in [16], using more than two wavelengths to measure the attenuation of light on the studied spectrum helps to highlight all the components of hemoglobin in the blood, and to obtain a pulse oximeter SpO_2 value that is considerably closer to the arterial blood's actual oxygen SaO_2 level. Thus, in present paper, a novel schematic of a pulse oximeter based on a multi-wavelength sensor is presented, which can help determining the other hemoglobin variants.

The basic idea of this prototype is to use as emitter a broadband visible and near-infrared (NIR) LED, such as SFH4737, which offers a 650 nm to 1050 nm spectral range emission in the interest zone (Fig. 2), and a multi-wavelength receiver sensor, both integrated in a transmittance-based system. In order to have a device well-suited for the general public, the main objective is to use low-cost components. In this case, the AS7263 spectral receiver, developed by AMS was considered a good choice. As shown in Fig. 3, this sensor has a 6-channel spectrometer in the red and near-infrared domain, which incorporates six independent optical filters. These six channels are strategically placed at the following wavelengths: 610 nm (R), 680 nm (S), 730 nm (T), 760 nm (U), 810 nm (V) and 860 nm (W). The main advantage of this approach, compared to the BQJ photodetector mentioned above, comes from a better separation between channels, each having a full-width half-max (FWHM) of 20 nm, as it can be seen in Fig. 3 [18].

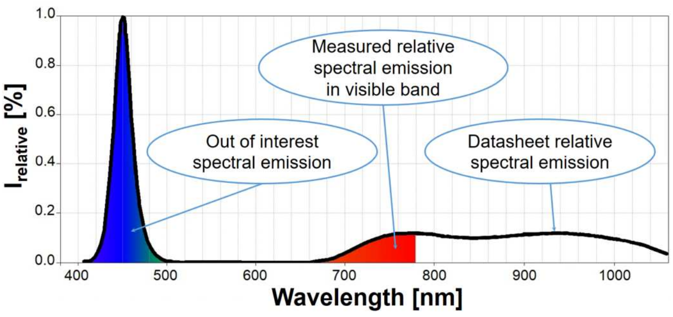

Fig. 2. Spectrum of the broadband visible & near-infrared LED SFH 4737, measured with a Sekonic C-800 Spectromaster color meter in the visible band, superimposed over the typical relative spectral emision from datasheet.

The proposed schematic is illustrated in Fig. 4. As opposed to Masimo Rad-57, where the eight wavelengths are used for HbCO and HbMet with the help of calibrated LEDs, but only the standard two wavelengths are used for SpO_2 measurements, this schematic can use a multiple-channel spectrum for a better SpO_2 reading accuracy. The component selection presented in this study will limit the use to the six channels mentioned above, because the basic idea is to demonstrate the proof-of-concept. The AS7263 is followed by an amplification block and a filter that rejects unwanted signals, including the blue component of the selected broadband NIR LED. The useful signals are then converted through the help of an analog-digital converter (ADC). The most important step is taking place in the data processing block, comprising a development board with a 32-bit ARM Cortex M4 processor at 180 MHz. Although it is out of the scope of this project, the captured signals can be compared with a calibration curve previously determined empirically with the help of healthy volunteers. Having more than two wavelengths at its disposal, this device can be calibrated to display the SpO_2 and PR values, but also SpMet and SpCO, which can help the estimation of the oxygen saturation with an increased accuracy.

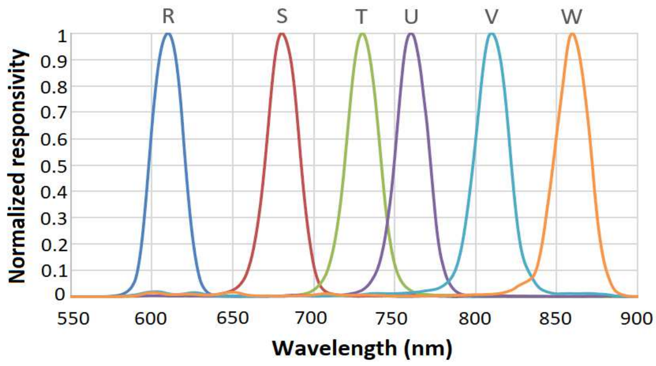

Fig. 3. Spectral responsivity of an AS7263.

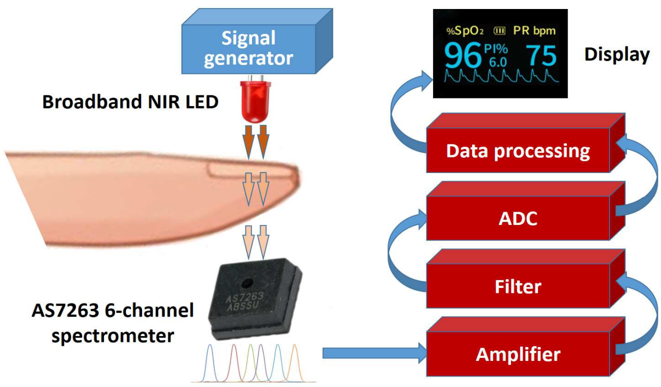

Fig. 4. Proposed pulse oximeter schematic.

IV. Materials and Methods

In order to hold the emitter and receiver devices in position, a 3D printed finger probe was designed, similar to a commercial one (Fig. 5). The incident light is produced by a broadband red and near-infrared LED, fitted in the upper part of the probe, whereas the 6-channel spectrometer is mounted in the lower part of the probe.

One of the authors participated voluntary in this experiment, which was conducted in two parts. The first part was dedicated to highlight the possibility to extract the pulse rate from the PPG signal acquired with the prototype, as a validation starting point for the main part of the experiment. The second part is basically a methods comparison study made as a proof-of-concept for this new low-cost concept of SpO_2 measuring in the current need for in-home monitoring. Thus, the new method of SpO_2 measuring is being evaluated by comparison with an established method of transmission with two wavelengths through human tissues.

Based on the results provided by the authors of [8], the Contec CMS50DL was chosen as a comparative low-cost device, being the one with the best results from a pool of six, in line with the FDA clearance threshold, i.e. an $A_{RMS} < 3\%$ for an actual oxygen saturation level SaO_2 from 70% to 100%. The Contec CMS50DL is a transmission mode device, which uses two wavelengths (i.e. 660 nm and 905 nm) that pass through human finger, and which are measured on the other end with a photodiode sensor.

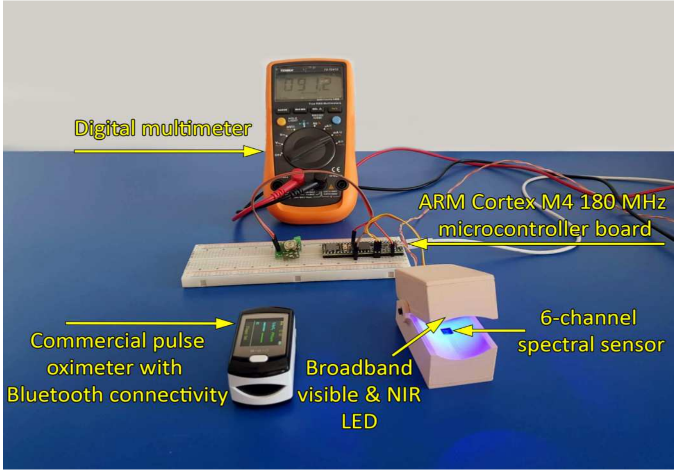

Fig. 5. Developed prototype and its testing setup.

A. *The Extraction Method of the Pulse Rate Value from the PPG Signal*

In order to determine the pulse rate, the prototype was placed on the volunteer's index finger and the values of the passing light were recorded in real-time. The number of samples was established at 1260 per minute, enough for an accurate PPG signal. For the incident light, the broadband red and near-infrared LED used has an emission power of 300 mW. It should be underlined here that, unlike the commercial available pulse oximeter, which has a total emission power of around 13 mW, the consumption of the prototype is rather high in this setup. The reason for using such a powerful emission is the low intensity of the LED in the red zone. Indeed, in the 610 nm channel, the irradiance is almost at the minimum for SFH 3747 (Fig. 2). However, the incident light can be further modulated with a PWM signal of a low duty-cycle, because the oxygen saturation has a relatively low variation. So, samples taken at predetermined intervals can be enough to keep track of how the oxygen saturation evolves, and that could decrease the average emission power to the levels seen in OTC devices. In order to counteract different finger thickness or skin tone, a pulse-amplitude modulation (PAM) can also be implemented, in conjunction with an automatic-gain control (AGC) function. Once the PPG signal shape is obtained, the pulse rate can be determined based on the peak detection method.

B. *The Measuring Method of the SpO_2*

The second part of the experiment started with measuring the reference values for all the six channel of the spectrometer sensor. The main task followed with data gathering from both the proposed device placed on the index finger, and the Contec device placed on the middle finger of the volunteer, while measurements were made in self-induced hypoxia. Because holding the breath in order to alter the SpO_2 by untrained individuals can't be maintained for more than two minutes [20], to have more time for settling the blood oxygenation the subject wore a breathing mask connected to a sealed bag of an 8-liter volume of air. Rebreathing the air from the bag leads to an O_2 decreasing and CO_2 increasing, which lowers the blood oxygen saturation level step-by-step. When the limit of tolerance was reached by the subject, the mask was removed, but the measurements were maintained until the SpO_2 was restored to a normal level.

V. Experimental Results and Discussions

In the first part of the experiment, the values obtained from the V-channel of the spectral sensor were recorded for about 60 seconds. The raw result was plotted in Fig. 6, where the cardiac rhythm is visible with the systolic and diastolic points. As it can be seen in the zoomed image, some of the cycles do not have a clear second wave, and the signal can be further filtered, but that should not be a problem for the peak detection stage. Another important aspect is that the signal shape must be inverted in order to have a real PPG visual, because the plotted data is for the attenuation of light, not for the blood pressure. After the peak detection step, the resulted pulse rate was calculated at 56 bpm, in line with the measurements made with the commercial device.

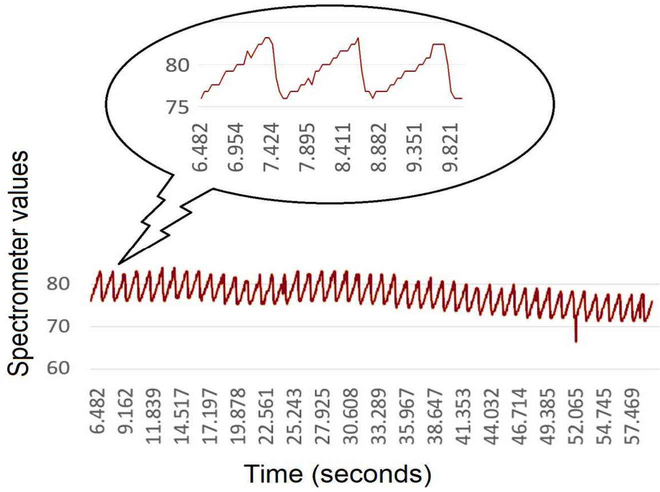

Fig. 6. Cardiac rhythm measured with the proposed pulse oximeter.

The experiment continued with the second part, the estimation of the SpO_2 level. At first, it was obvious that the first two channels (R and S) have very weak signals, due to the low intensity of LED light in this region of the spectrum. In order to take advantage of these channels, pulse modulation with an increased power of emission could be as well the solution, but in this experiment, only the last four channels were considered, and R and S channel were kept as references.

The volunteer had at the beginning of the measurements an oxygen saturation level of 98%, and reached a 92% SpO_2 level in about five minutes of rebreathing in the bag, without experiencing disagreeable symptoms or major change in vitals. After the discontinuation of carbon dioxide inhalation, the SpO_2 level increased to normal in less than one minute, reaching a normal SpO_2 level of 98%, as seen in Fig. 7. After data gathering from all the channels, having the maximum and minimum values of oxygen saturation, calculations were made based on (5), to estimate the values of the parameters *a* and *b*. Having the raw data for the estimated blood oxygen saturation, a ten-second period moving average was plotted, in order to see the trending lines. From all four channels considered, the best results offered the most distanced channels, T and W. The recorded values from Contec CMS50DL pulse oximeter, pictured in the upper part of the figure, show a direct correlation with the trending line of the estimated SpO_2, which demonstrates the viability of the concept. The graphic also shows that the commercial pulse oximeter displayed the SpO_2 values with a few seconds delay, which is in line with what was seen during testing. The estimated values for the oxygen saturation, based on (5), are consistent with the values measured with the comparison commercial device, but a more accurate calibration is necessary to be empirical determined with the help of healthy volunteers, under medical supervision, until a blood oxygen saturation of 70% is reached.

Based on the results collected from the remaining channels, other variations of hemoglobin can also be determined, allowing for an increased accuracy in SpO_2 estimation. Future improvements focused on signal filtering and elimination of the motion artifacts.

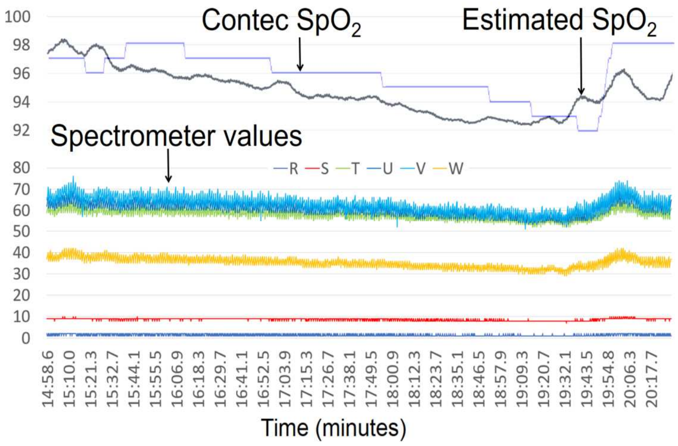

Fig. 7. Recorded values from Contec CMS50DL commercial pulse oximeter and the estimated SpO_2 representation superimposed over the recorded values from AS7263 6-channel spectrometer.

VI. Conclusion

A novel method for pulse oximetry estimates has been introduced. This paper provides a novel schematic based on a broadband NIR LED, such as SFH4737, and a 6-channel multi-wavelength sensor, such as AS7263, capable of an accurate detection of SpO_2, but can also be adapted to estimate hemoglobin derivatives, such as SpMet, or SpCO. Unlike other solutions presented in various research papers, this new prototype uses only one broadband red and near-infrared LED for incident light emission, and a low-cost 6-channel spectral sensor with a full-width half-max (FWHM) of 20 nm, that allows a good separation between channels for better comparisons across the spectrum. Although due to safety reasons the certain certification investigations could not be performed, this approach offers the benefits of improved accuracy and enhanced precision with respect to the classical methods when abnormal levels of hemoglobin variants are present. On the downside, the current setup has a higher power consumption, but this issue can be addressed by modulating the incident light with a PWM signal of a low duty-cycle.

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