A Novel Calibration-Free Method of Measurement of Oxygen Saturation in Arterial Blood

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Abstract—In present-day pulse oximeters, oxygen saturation in arterial blood ($\text{SpO}_2$) is computed by utilizing an empirical relationship extracted from a calibration curve. The calibration curve is obtained by curve-fitting data acquired from volunteers. A novel method of computation of $\text{SpO}_2$ that does not require the use of a calibration curve is presented in this paper. Based on a model for the attenuation of light through skin, tissue, bone, and blood, suitable processing steps are identified so that the analytical expression derived for the estimation of $\text{SpO}_2$ becomes free of not only patient but sensor-dependent parameters as well. The experimental results presented in this paper establish the efficacy of the proposed method.

Index Terms—Biomedical signal processing, oxygen saturation, photoplethysmograph (PPG), pulse oximeter.

I. INTRODUCTION

A part from necessary nutrients, arterial blood carries oxygen required so that the body functions properly. Hence, for the assessment of the health of the cardiopulmonary system of a patient, a physician requires information on the amount of oxygen being carried by the patient’s arterial blood [1]. The level of oxygen in arterial blood is indirectly determined by measuring the oxygen saturation ($\text{SaO}_2$) in arterial blood. If the concentrations of oxygenated hemoglobin (oxyhemoglobin) and hemoglobin (also known as deoxy or reduced hemoglobin) in arterial blood are represented by $\langle \text{HbO} \rangle$ and $\langle \text{Hb} \rangle$, respectively, then

$$\text{SaO}_2\% = \text{SpO}_2\% = \frac{\langle \text{HbO} \rangle}{\langle \text{Hb} \rangle + \langle \text{HbO} \rangle} \times 100 = \frac{Q}{1 + Q} \times 100 \quad (1)$$

where $Q = \langle \text{HbO} \rangle/\langle \text{Hb} \rangle$. Equation (1) is valid if and only if the levels of dyshemoglobin in arterial blood, namely, carboxyhemoglobin and methemoglobin, are negligible. Pulse oximetry is a popular and noninvasive technique for the measurement of $\text{SaO}_2$. When $\text{SaO}_2$ is measured by using the principle of pulse oximetry, it is expressed as $\text{SpO}_2$ [2]. In pulse oximetry, a part of the body is illuminated with two monochromatic light sources that have wavelengths in the red and infrared (IR) regions. Either the reflected or transmitted light from or through the body is detected to obtain red and IR photoplethysmographs (PPGs) [3]. The source and the detector have to be housed on the same plane, as shown in Fig. 1(a), to get a PPG utilizing the reflected light ($\text{PPG}_R\lambda$, where $\lambda$ is the wavelength of light). To obtain a PPG employing the transmitted light ($\text{PPG}_T\lambda$), the source and the detector have to be arranged on two different but parallel planes, as shown in Fig. 1(b).

Fig. 1. PPG sensor heads. (a) Reflectance type. (b) Transmission type.

Fig. 2. Components of a typical PPG signal.

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portion (≈ 0.5%) of the received light (reflected/transmitted) is due to attenuation by arterial blood.

Under normal physiological conditions, the arterial blood is 97% saturated with oxygen. A healthy non-smoking person should have SpO₂ between 94% and 100%. Patients undergoing surgery or recovering from surgery or admitted into intensive care units may have oxygen saturation in the range of 90%–100%. In fact, SpO₂ below 90% in adults and children will quickly lead to life-threatening complications and will need immediate clinical intervention. It has been shown that SpO₂ values very rarely go below 70%, and hence, a measurement of SpO₂ below 70% is rarely needed in adults [8], [9]. Thus, conventional pulse oximeters are accurate (within 2%) in the 80%–100% range of SpO₂ [2]. On the other hand, pulse oximeters intended for monitoring the oxygen status of a fetus during labor and delivery should work in the range of 15%–65% [10], [11]. As a conventional pulse oximeter tends to read less accurately at low saturations, fetal pulse oximetry has to be addressed using special probes and enhanced calibration algorithms. A recent study revealed that the accuracy of SpO₂ estimation depends on the wavelength used in the pulse oximetry and concluded that, for high oxygen saturations (child and adult SpO₂ > 80%), the choice of 660- and 900-nm band emitters appears optimal while 735- and 900-nm band emitters perform better at low saturations (fetal pulse oximetry SpO₂ of 20%–70%) [8].

Traditional methods compute the ratio of ratios R as

\[ R = \frac{AC_R/DC_R}{AC_{IR}/DC_{IR}} \]

where ACₕ and DCₕ represent the magnitudes of the pulsatile and dc parts, respectively, of the PPG obtained at the red wavelength. Similarly, ACₜ and DCₜ are the magnitudes of the pulsatile and dc portions, respectively, of the IR PPG. SpO₂ is then calculated by employing an empirical equation

\[ \text{SpO}_2 = K_1 + K_2 R \]  

where \( K_1 \) and \( K_2 \) are constants extracted from a calibration curve obtained by curve fitting data acquired from volunteers [2]–[7].

In conventional pulse oximetry, the empirical calibration equation for SpO₂ estimation is derived from the data obtained from a large group of healthy non-smoking young volunteers. This results in a pulse oximeter that relies upon fixed calibration curves to predict SpO₂ from measured PPG signals [10]. Consequently, each pulse oximeter manufacturer will have their own calibration curves based on the group of volunteers tested [12], [13]. In other patented algorithms [14], [15], an instantaneous ratio \( R' \) is found from the ratio of the derivatives of red and IR intensities, which is again used in an empirical formula to compute the average SpO₂ value. Recently, a calibration-free method, based on two wavelengths that are close to each other, has been reported but again utilizes a slightly modified ratio \( R \) [16]. In all the patents and the information available in the literature, the ratio of ratios \( R \) (that is taking the AC/DC values of red and IR PPGs) and an empirical calibration equation are still used for SpO₂ computation.

A novel method of evaluation of SpO₂ suitable for children and adults that does not rely on a calibration curve was proposed earlier by the authors [17]. In this paper, the method is analyzed and experimentally verified. The analytical expression derived for the computation of SpO₂ in the method being presented in this paper is free of not only patient-reliant parameters but sensor-dependent parameters as well. In the method proposed in this paper, it has been shown for the first time, theoretically and by way of suitable experimentation, that the application of natural logarithm removes the influence of source intensity, detector sensitivity, and patient-dependent parameters from the ac part of the PPG signals. It is also demonstrated that, after applying natural logarithm, the parameters of the ac part alone are sufficient to compute SpO₂ values without the need for any calibration constants derived from data obtained from volunteers.

II. PROPOSED SLOPE-BASED METHOD

The proposed method of measurement of SpO₂ is applicable for both PPGRₜ and PPΓTₜ. However, to ensure brevity, explanations and derivations are given here only for a transmission-type PPG. One can easily obtain relevant expressions and explanations that are applicable for a PPGRₜ simply by replacing the word “absorption” with “reflection.” Fig. 2 shows a typical PPG and its components. Majority of the photons emitted by the red and IR sources of the sensor head in Fig. 1(b) pass through path 1 through the finger, which is made of epidermis–tissue–soft bone–tissue–epidermis, and reach the detector. Since the attenuation in this path does not vary with time, the component in the PPG (output of the detector) due to the attenuation in path 1 will be a dc voltage, as shown in Fig. 2. Only a very small number of photons emitted by the sources go through the path, namely, path 2 that includes veins (resulting in a very low frequency component at the output of the detector), and a much smaller number of photons go through the path 3, namely, epidermis–tissue–arteries–soft bone–tissue–epidermis, resulting in a detected pulsatile ac voltage at the heart rate [2]. Hence, a typical PPG is made of a dc, low frequency, and pulsatile ac (at the heart rate) components as shown in Fig. 2.

To model the propagation of light from the source to the detector in Fig. 1(b), the path of light (intensity \( I_{\lambda}(l) \) from the source is assumed to be cylindrical in shape, having a uniform cross section of area \( A \) and length \( T_F \) [thickness of the finger in Fig. 1(b)], as shown in Fig. 3. Nearly 90% of the volume of the path is made of cells of dermis, tissue, and bones. A disk (in the path \( l = 0 \) to \( T_F \)) of thickness \( dl \) having an attenuation \( di \) across it, on an incident light intensity \( i_l \), is considered for analysis. The composition of the cells in the disk depends

![Fig. 3. Model showing the light path through the finger in Fig. 1(b).](image-url)
The attenuation of light, depending on its optical property, and scatters some of its intensity on the position of the disk. Each cell in the disk attenuates light, as shown in Fig. 4(a), is now modeled as a fraction of the area, for example, $\sigma_{TA}$ of the cross-sectional area $A_T$, of that cell is completely opaque and the rest of the cross-sectional area of that cell $(A_T - \sigma_{TA})$ is completely transparent, as shown in Fig. 4(b). Then, the opaque area within the disk due to all tissue cells in the disk is $\sigma_{TA} NT_{Adl}$, where $NT$ is the cell concentration of tissue cells (cells per unit volume) within $dl$. The total opaque area due to all cells is

$$\sigma_{DA} NT_{Adl} + \sigma_{TB} NT_{Adl} + \sigma_{BA} NT_{BAdl} = (\sigma_{DA} NT_{D} + \sigma_{TB} NT_{T} + \sigma_{BA} NT_{B}) Adl.$$  

Here, $\sigma_{DA}$ and $\sigma_{BA}$ are the equivalent opaque areas representing the attenuation at the wavelength $\lambda$, and $ND$ and $NB$ are the cell concentrations of the dermis and bone cells, respectively. Then, the attenuation $di_l$ across the disk of length $dl$ is

$$di_l = -\frac{\text{Total opaque area}}{\text{Total area}} il.$$  

Rearranging and integrating (3) with limits $l = 0$ to $l = T_F$, we get

$$\int_{l=0}^{l=T_F} di_l = \int_{l=0}^{l=T_F} -(\sigma_{DA} NT_{D} + \sigma_{TB} NT_{T} + \sigma_{BA} NT_{B}) dl$$

$$\ln(il)|_{l=T_F} = -(\sigma_{DA} NT_{D} + \sigma_{TB} NT_{T} + \sigma_{BA} NT_{B})|_{l=0}. $$

On simplification, we get

$$\ln \left( \frac{I_{0\lambda}}{I_{\lambda}} \right) = -(\sigma_{DA} NT_{D} + \sigma_{TB} NT_{T} + \sigma_{BA} NT_{B})T_F.$$  

Here, $I_{0\lambda}$ is the intensity of the light received at the detector, and $I_{\lambda}$ is the intensity of the source emitting light at a wavelength $\lambda$. Applying natural logarithm to (4) results in

$$I_{0\lambda} = I_{SA} e^{-(\sigma_{DA} NT_{D} + \sigma_{TB} NT_{T} + \sigma_{BA} NT_{B})T_F}.$$  

The light intensity $I_{0\lambda}$, as given in (5), falling on the photodetector will give rise to a dc component at the output of the photodetector. The dc part $V_d = v_{dc\lambda}|dc$ of the output $v_{dc\lambda}$ of the photodetector having a sensitivity $K_{D\lambda}$ is

$$V_d = K_{D\lambda} I_{0\lambda} = K_{D\lambda} I_{SA} e^{-(\sigma_{DA} NT_{D} + \sigma_{TB} NT_{T} + \sigma_{BA} NT_{B})T_F}.$$  

It is customary to express concentrations of constituent parts in terms of mol·L$^{-1}$. In that case, the opaque area $\sigma$ must be multiplied by a factor $((6.023/2.303)10^{20}$), giving rise to the extinction coefficient $\varepsilon$ (L·mol·cm$^{-1}$). With concentrations expressed in mol·L$^{-1}$, (6) becomes

$$V_d = K_{D\lambda} I_{0\lambda} = K_{D\lambda} I_{SA} e^{-(\sigma_{DA} NT_{D} + \sigma_{TB} NT_{T} + \sigma_{BA} NT_{B})T_F}.$$  

Here, $\varepsilon_{DA}$, $\varepsilon_{TB}$, and $\varepsilon_{BA}$ are the extinction coefficients, and $\langle D \rangle$, $\langle T \rangle$, and $\langle B \rangle$ are the molar concentrations of epidermis, tissue, and bone cells, respectively.

As mentioned earlier, some of the photons also travel through arteries carrying blood that contains 55% of plasma, 43% of red blood cells (mostly made of $Hb$ and $HbO$), 1.5% of white blood cells (leukocytes), and 0.5% of platelets. Of these, attenuation due to red blood cells alone is significant as the amounts of white blood cells and platelets are negligible compared to the number of red blood cells while plasma has nearly zero optical attenuation at the red and IR wavelengths. To model the path of light that also traverses through arteries, all the individual arteries that are in the path of light are considered as a single equivalent artery, having a height of interaction with light $H$ and width $x$, as shown in Fig. 3. As blood is pumped into the arteries, the vessels enlarge and contract at the rate set by the heart, and hence, $x$ is a time-varying quantity. Let $\hat{x}$ be the value of $x$ when the blood flowing in the equivalent artery is a maximum. Then, the attenuation $di_{\lambda}$ due to arterial blood is

$$di_{\lambda} = -i_{\lambda} \frac{\varepsilon_{Hb\lambda}(Hb) + \varepsilon_{HbO\lambda}(HbO)}{\hat{x}H} dx.$$  

Here, $i_{\lambda}$ is the intensity of that part of the input light that also passes through arterial blood vessels, i.e., the intensity of light being incident on the single equivalent artery having a width $x$ and height $H$. $\langle Hb \rangle$ and $\langle HbO \rangle$ are molar concentrations, and $\varepsilon_{Hb\lambda}$ and $\varepsilon_{HbO\lambda}$ are the well-known extinction coefficients of hemoglobin and oxyhemoglobin, respectively [10]. Rearranging (8) and integrating over the limit $x = 0$ to $x = \hat{x}$ results in

$$\ln \left( \frac{i_{\lambda}}{i_{\lambda}} \beta I_{0\lambda} \right) = \left( \frac{\varepsilon_{Hb\lambda}(Hb) + \varepsilon_{HbO\lambda}(HbO)}{\hat{x}} \right) \left[ x = \hat{x} - x = 0 \right]. $$

Here, $\beta I_{0\lambda}$ is the small fraction of light ($\beta \approx 0.005$) from the source $\langle \lambda_s \rangle$ that is also attenuated by the arteries. Hence, the pulsatile component $v_{dc\lambda}|ac$ at the output of the detector is

$$v_{dc\lambda}|ac = v_{dc\lambda} = K_{D\lambda} \beta I_{0\lambda} e^{(\varepsilon_{Hb\lambda}(Hb) + \varepsilon_{HbO\lambda}(HbO)) x^{\hat{x}}}_{x=0}. $$

(9)
Applying natural logarithm to (9), we get

\[
\ln(v_{0\lambda}) = \ln K_{D\lambda} I_{0\lambda} - \left( \frac{\langle Hb\lambda (Hb) + \varepsilon_{HbO\lambda} (HbO) \rangle x^2}{x = \hat{x}} \right) |_{x = 0}.
\]

Here, \(\ln(K_{D\lambda} I_{0\lambda})\) is dependent on the patient and the sensor parameters (intensity of source and sensitivity of the detector) but is a dc component (value varies for different patients but is not a time-varying quantity). The pulsatile portion of (10) is

\[
\ln(v_{0\lambda})_{\text{pulse}} = v_{\lambda} = \frac{\langle Hb\lambda (Hb) + \varepsilon_{HbO\lambda} (HbO) \rangle x^2}{2\hat{x}}.
\]

Obtaining \(V_{p\lambda}\), which is the peak value (when \(x = \hat{x}\)) of \(v_{\lambda}\), from (10) and substituting it in the same equation results in

\[
v_{\lambda} = \frac{\langle Hb\lambda + \varepsilon_{HbO\lambda} Q\rangle^2 (Hb)^2 x^2}{4V_{p\lambda}}.
\]

Here, \(Q = \langle HbO \rangle / \langle Hb \rangle\). During the period \((t_1 - t_2)\) where \(\langle Hb \rangle^2 x^2\) linearly varies with time, for example, between \(x_1\) and \(x_2\), (11) can be expressed as

\[
v_{\lambda} = m_{\lambda} t_{l_{t_2}}^{t_{t_1}}
\]

where \(m_{\lambda}\) is the slope of \(\langle Hb\lambda + \varepsilon_{HbO\lambda} Q\rangle^2 (Hb)^2 x^2 / 4V_{p\lambda}\).

In the proposed method, natural logarithms of the red and IR PPG signals \(v_{0R}\) and \(v_{0IR}\) are first evaluated, and the pulsatile portions of \(\ln(v_{0R})\) and \(\ln(v_{0IR})\) are extracted to obtain \(v_{R}\) and \(v_{IR}\), respectively. The peak-to-peak values \(V_{pR}\) and \(V_{pIR}\) of \(v_{R}\) and \(v_{IR}\), respectively, are then determined. The linear portions of \(v_{R}\) and \(v_{IR}\) are identified, and the slopes \(m_{R}\) and \(m_{IR}\), respectively, are computed, where

\[
m_{R} = \frac{\langle HbR + \varepsilon_{HbOR} Q\rangle^2 (Hb)^2 x^2}{4V_{pR}}
\]

\[
m_{IR} = \frac{\langle HbIR + \varepsilon_{HbOIR} Q\rangle^2 (Hb)^2 x^2}{4V_{pIR}}.
\]

Here, \(\varepsilon_{HbR}\) and \(\varepsilon_{HbIR}\) are the extinction coefficients of \(Hb\) and \(HbO\), respectively, at the chosen red wavelength, and \(\varepsilon_{HMR}\) and \(\varepsilon_{HMR}\) are the extinction coefficients of \(Hb\) and \(HbO\), respectively, at the selected IR wavelength. \(Q\) is then obtained by dividing (12) by (13) as

\[
Q = \frac{m_{IR} V_{pIR} - m_{R} V_{pR}}{m_{IR} V_{pIR} - m_{IR} V_{pIR}}
\]

Substituting the value of \(Q\) from (14) in (1) results in (15), as shown at the bottom of the page.

It is easily seen from (15) that the expression for the computation of \(\text{SpO}_2\) is independent of patient-dependent variables as well as the intensities of the red and IR sources and the sensitivities of the red and IR detectors. A comparison of (2) employed for the computation of \(\text{SpO}_2\) in traditional pulse oximeters and (15) employed in the present method reveals the following.

1. For the first time, \(\text{SpO}_2\) is computed without resorting to the ratio of ratios \(R\) and an empirical calibration equation.
2. Hence, the proposed method does not require calibration constants \(K_1\) and \(K_2\) [see (2)]. Thus, the computed value of \(\text{SpO}_2\) is not influenced by the composition of volunteers employed for obtaining calibration constants \(K_1\) and \(K_2\).
3. The parameters of the ac part (slope and peak-to-peak value) of the PPG after applying natural logarithm alone are sufficient to compute \(\text{SpO}_2\), and the dc part (patient and sensor dependent) does not influence the computation of \(\text{SpO}_2\).

The details of the experiments conducted to validate (15) and the prototype built and tested are given next.

### III. EXPERIMENTAL RESULTS

To ascertain the feasibility of the proposed scheme, a prototype was built and tested. The details of the prototype and the testing procedure adopted are given next. The computation of \(\text{SpO}_2\) in the proposed method, per (15), requires the slopes \((m_{IR} \text{ and } m_{R})\) and peak-to-peak amplitudes \((V_{pR} \text{ and } V_{pIR})\) of red and IR PPG signals. Hence, the experiments are conducted so as to compare the slopes and peak-to-peak values at different source intensities and detector sensitivities.

#### A. Materials

A clip-on sensor, as shown in Fig. 1(b), was developed, housing a red LED (emitting light at 660 nm) and an IR LED (emitting light at 940 nm) on one side and a photodiode detector on the other side of a soft plastic clip. The red and IR LEDs and the photodiode are connected to the signal conditioning circuit shown in Fig. 5. For brevity, the circuit schematic of the IR channel alone is shown in Fig. 5, and the red-channel circuit schematic is identical to that of the IR channel. The IR LED is controlled by a voltage-to-current converter comprising opamp OA1, transistor TR1, and resistor \(R_D\). The current through the IR LED is set to a particular value by controlling the switch SW1. With the control lines \(A1A0 = 00\), switch SW1 will be in position 1, and the current \(I_D\) through the IR LED will be \(I_D = (V_Z / R_D)A\). When \(A1A0 = 01\), SW1 is set at position 2, and \(I_D = (V_Z (R_2 + R_3 + R_4) / R_D (R_1 + R_2 + R_3 + R_4))A\). Similarly, when SW1 is set in position 3, \(I_D = (V_Z (R_3 + R_4) / R_D (R_1 + R_2 + R_3 + R_4))A\), and with SW1 set in position 4, \(I_D = (V_Z (R_1) / R_D (R_1 + R_2 + R_3 + R_4))A\). The values of \(V_Z\) and the resistances \(R_1 - R_4\) are chosen such that a current of 8 mA flows through the LED when switch SW1 is in position 1, 6 mA when in position 2, and 4 and 2 mA when switch SW1 is set in positions 3 and 4,

\[
\text{SpO}_2\% = \frac{\varepsilon_{HbR} \sqrt{m_{IR} V_{pIR}} - \varepsilon_{HMR} \sqrt{m_{R} V_{pR}}}{\sqrt{m_{IR} V_{pIR} (\varepsilon_{HbR} - \varepsilon_{HbR})} - \sqrt{m_{R} V_{pR} (\varepsilon_{HMR} - \varepsilon_{HMR})}}
\]
The photodiode converts the intensity of the light being detected by it to a proportional current. The photodiode current is converted to a proportional voltage by employing an $i$-$v$ converter made of opamp OA2 and $R_{i}$. The dc part of the output of OA2 is extracted by the low-pass filter realized with opamp OA3, and the pulsatile ac part is extracted by opamp OA4. The PPG signal is separated into dc and ac parts as the gain required to amplify the dc part is quite different from the gain required to amplify the ac portion. The dc and ac parts of the IR PPG signals thus obtained from the photodiode are acquired by using a 16-b data acquisition card, i.e., DAQPad-6015 manufactured by National Instruments, connected to the universal-serial-bus port of a personal computer. Suitable software is written in the LabVIEW virtual instrumentation environment for acquiring and processing the samples of red and IR PPG signals. The switches SW1 and SW2 (of the red channel, which is not shown in Fig. 5) are also controlled through the digital I/O lines of DAQPad-6015. Although a patient being tested will come into contact only with the optical output of the sensor, all the equipment utilized were connected to the utility power through an isolation transformer, and the DAQ system was run on battery to ensure absolute safety to the patients.

**B. Experimentation**

Experiments were conducted to demonstrate the following.

I) The pulsatile portions of the red and IR PPG signals, after applying natural logarithm, as given in (11), are independent of the following:

A) the intensities of the red and IR sources;
B) the detector (red and IR) sensitivities;
C) patient-dependent parameters.

I) The computation of SpO$_2$ as per (15) is independent of the following:

a) the intensities of the red and IR sources;

b) the detector (red and IR) sensitivities;

c) patient-dependent parameters.

To demonstrate that applying natural logarithm to a PPG removes the influence of the source intensity, the currents through the red and IR LEDs were varied over the range of 2–8 mA in steps of 2 mA by changing the positions of switches SW1 and SW2 every millisecond as 1, 2, 3, 4, 1, 2, 3, 4, 1, 2, . . . . This process changes the current through the red and IR LEDs, cyclically as 8, 6, 4, and 2 mA, resulting in the intensity variations as 12, 9, 6, and 3 mcd. Data pertaining to every millisecond were acquired at the middle of the 1-ms period. Since the acquisition time was 5 $\mu$s and the rise and fall times of the switches were 1 $\mu$s each, the effect of transients due to switching is completely eliminated. The resulting dc and ac parts of the red and IR PPG signals for each intensity value are recorded for 20 volunteers. Informed consents were obtained from the volunteers, and the entire procedure was approved by the ethics committee of the Indian Institute of Technology (I.I.T.) Madras. The data pertaining to a particular intensity, for example, 12 mcd, were grouped together by concatenating samples 1, 5, 9, 13, . . . . Thus, the acquired PPG signals at the four intensity levels are first separated and plotted. It should be noted here that this process obtains four different sets of PPG signals (red and IR) of widely differing intensity levels within one PPG cycle. Traces (a)–(d) of Fig. 6 show a typical plot of IR PPG signals for the four different source intensity levels. Then, natural logarithm is applied to the PPG signals shown in traces (a)–(d), and the pulsatile portions of the resulting signal alone are extracted and shown in trace (e). It is evident from Fig. 6 that, even though the intensity of the source has increased over fourfold, the peak-to-peak amplitude values and slopes of the processed PPG signals have remained constant. The very small deviations seen in traces of Fig. 6(e) are due to small changes in the scattering of light as the source intensity increases, increasing $\hat{x}$. Thus, $V_{p,\lambda}$ slightly varies with the intensity of the source within a PPG cycle, resulting in very small variations between the processed PPG signals obtained at different intensities, as shown in Fig. 6(e). The cycle-to-cycle amplitude variations in PPG shown in Fig. 6 are due to breathing and random blood-flow variations in the microvessels. It has been shown that, using these variations, breathing patterns and breath rate can also be derived from PPG signals. Table I lists the measurements.
made on the PPG signals obtained with the four source intensity levels. It is evident from Fig. 6 and Table I that applying the natural logarithm removes the influence of intensity on the pulsatile portion of the processed signal, and (11) is thus validated.

Then, the intensities of the red and IR LEDs were set at 9 mcd (6-mA LED current) each, and PPG signals were obtained with four different detector sensitivity settings (by varying the gain of the particular channel of the DAQPad 6015). Table II lists the peak-to-peak values and the slopes of a typical raw PPG and after applying natural logarithm to the four different sets (corresponding to the four different sensitivity settings) of red and IR PPG signals. It is evident from Table II that applying the natural logarithm delineates the influence of detector sensitivity on the pulsatile portion of a PPG, apart from the effect of source intensity.

During the recordings of the PPG signals with variations in the source intensity levels and also with different detector sensitivities, the SpO\textsubscript{2} values of the volunteers (indicated by a commercial pulse oximeter (CPO), model Planet 50, manufactured by L&T Medical systems [18], Mysore, India) were noted down. The SpO\textsubscript{2} value for each recording was also computed by using (15). To evaluate (15), the slopes and peak values are obtained from the processed red and IR PPGs, and the required values of the extinction coefficients $\varepsilon_{HbR}$, $\varepsilon_{HbIR}$, $\varepsilon_{HbOR}$, and $\varepsilon_{HbOIR}$ were obtained from published literature [19], which were shown in Fig. 7. Table III portrays the SpO\textsubscript{2} values computed by using the proposed method for five volunteers at two source intensity levels and the readings that were simultaneously recorded with a CPO. Since the oxygen saturation level is a dynamic value, the average of several SpO\textsubscript{2} readings and the standard deviation obtained are listed in Table III. As all the volunteers were healthy subjects, the SpO\textsubscript{2} values were in the range of 97%–98%. Experiments for lower values of SpO\textsubscript{2} were not carried out due to lack of medical facility for administering hypoxic gases to volunteers to reduce their SpO\textsubscript{2} levels. However, upon request, L&T Medical systems [18] provided red and IR PPG data for different oxygenation conditions ranging from 85% to 100% of SpO\textsubscript{2}. The proposed method was applied on these data, and the estimated SpO\textsubscript{2} values rounded off to the nearest integer are listed in Table IV.

Fig. 6 and the readings in Tables I–IV validate the practicality of the proposed method for different saturation values.
IV. CONCLUSION

A novel calibration-free method of measurement of oxygen saturation in arterial blood (SpO$_2$) has been proposed. Based on the model proposed in this paper for the attenuation of light in a typical PPG sensor, appropriate processing steps are derived to extract an analytical expression for the computation of SpO$_2$ by utilizing a couple of PPG signals: one obtained in the red wavelength and the other at the IR wavelength. The processing steps are chosen such that SpO$_2$ is directly determined by employing measurements made on the processed red and IR PPG signals and the well-established extinction coefficients of hemoglobin and oxyhemoglobin, without resorting to calibration constants derived from data obtained from volunteers. The advantage of the proposed method lies in the fact that the analytical expression derived for the computation of SpO$_2$ is free of patient- and sensor-dependent parameters. Since the proposed method does not rely on an empirical calibration equation, it does not require extensive calibration, such as that required in present-day pulse oximeters. The prototype built and tested establishes the practicality of the proposed method.

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REFERENCES


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