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A PULSE OXIMETER FOR MEASURING THE BLOOD OXYGENATION LEVEL CONSIDERING THE CARBOXYHEMOGLOBIN CONCENTRATION: PRINCIPLES OF DEVELOPMENT, COMPUTER MODEL AND ACCURACY ASSESSMENT

The main method for estimating the level of arterial blood oxygenation is pulse oximetry, which has the advantages of being fast, simple, reliable, and non-invasive. However, in well-known pulse oximeters, oxygen saturation is determined only by hemoglobin functional fractions, which reduces the device accuracy and is unacceptable in certain clinical cases. The known pulse oximeter improvement that considers the dysfunctional fraction concentration, especially carboxyhemoglobin, when measuring the level of blood oxygenation is an actual scientific and technical task. ***The research subjects.*** *Mathematical, algorithmic, and technical support of a pulse oximeter that measures blood oxygenation levels considering the carboxyhemoglobin concentration.* ***Objective.*** *To expand the pulse oximeter functionality to consider the concentration of carboxyhemoglobin in arterial blood.* ***Methods.*** *Methods of computer simulation for developing a model and estimating the pulse oximeter accuracy that measures the blood oxygenation level considering the carboxyhemoglobin concentration.* ***Results.*** *The theoretical statements of measuring the level of blood oxygenation considering the carboxyhemoglobin concentration and the simplest pulse oximeter structural diagram for measuring are developed. An additional LED used in the pulse oximeter is proposed, and the wavelength choice is justified on the condition of maximizing the carboxyhemoglobin contribution to the optical density of the biological object. Computer models of a traditional pulse oximeter and a pulse oximeter with an additional LED were developed, simulation research was conducted using the developed models, and the device accuracy for measuring the level of blood oxygenation was estimated considering the carboxyhemoglobin concentration.* ***Conclusions.*** *Simulation studies based on the developed models show that the proposed pulse oximeter, compared with the known one, allows determining and estimating a decrease in blood oxygenation caused by the carboxyhemoglobin concentration increasing in the patient's blood. Considering that light is also absorbed by the third derivative of hemoglobin, carboxyhemoglobin, increases the accuracy of the proposed pulse oximeter in measuring functional saturation.*

Keywords: *pulse oximetry; oxygenation; carboxyhemoglobin; development; computer model; saturation; hypoxemia; accuracy.*

1. Introduction

1.1. Motivation.

There are human body functioning pathologies that are accompanied by hypoxemia, and the patient's blood oxygenation level requires constant monitoring.

Pulse oximetry is the main technique for arterial blood hemoglobin oxygen saturation continuous estimation [1, 2]. The advantages are speed, simplicity, reliability, and non-invasiveness in determining low blood oxygen levels before clinical features become noticeable. A medical device designed to measure oxygen concentration in the blood is a pulse oximeter (PO). Portable POs powered by primary galvanic elements are available for transport and home monitoring of blood oxygenation [3].

The importance of monitoring the oxygen saturation is increasing significantly in the current COVID-19 pandemic conditions [4], when patients have significant lung tissue damage. Oxygen saturation measured by pulse oximetry is higher in hypoxemia than the true oxygen saturation measured invasively, thereby increasing the risk of occult hypoxemia [5]. Therefore, improving the accuracy of known POs, and developing devices for measuring blood oxygenation levels in patients with pathological conditions, is an actual task.

1.2. State of the art

In well-known POs, the saturation of arterial blood hemoglobin with oxygen is determined by equation [6]:

$$SpO_2 = \frac{C_{HbO_2}}{C_{HbO_2} + C_{HbH}}, \quad (1)$$

where C_{HbO2} is the oxyhemoglobin concentration, mol/l; C_{HbH} is the deoxyhemoglobin concentration, mol/l.

SpO_2 value is the functional saturation. The numerical value shows the arterial blood hemoglobin proportion that is combined with oxygen (oxyhemoglobin,

HbO_2) of the functional hemoglobin total concentration that can take part in oxygen transport.

SpO_2 determination in POs is carried out by the optical two-spectral method [7], and two LEDs are usually used as a light source: red and infrared with approximate emitted wavelengths of $\lambda_1 = 660$ and $\lambda_2 = 940$ nm, respectively.

Monochromatic light with wavelength λ_j and intensity $I_{0,j}$ is partially absorbed and travels through the arterial blood. The light intensity at the output is I_j as determined by the Bouguer-Lambert-Beer law:

$$\lg \frac{I_{0,j}}{I_j} = l \cdot \sum_{i=1}^{n} \left(\varepsilon_{j,i} \cdot C_i \right), \quad (2)$$

where C is the hemoglobin derivative molar concentration in the blood, mol/l; l is the arterial blood layer thickness, cm; n is the number of hemoglobin derivatives in the blood; ε is the extinction coefficient, l/(mol·cm). The following indexes are used in the formula: i is the hemoglobin derivative serial number and j is the light wavelength serial number.

In (1), oxygen saturation is determined by two derivatives of hemoglobin, so n in (2) is 2.

However, the arterial blood layer thickness l changes due to changes in the blood circulation in vessels that are done periodically and are determined by the pulse. In addition, the emitted light is absorbed not only by arterial but also by venous blood along with the skin [8].

Therefore, an auxiliary value R is introduced into the saturation calculation, which is determined by the relative change in each wavelength light intensity (Fig. 1) travelled through the body tissue during one cardiac cycle:

$$R = \frac{\lg\left(\frac{I_{min,1}}{I_{max,1}} \right)}{\lg\left(\frac{I_{min,2}}{I_{max,2}} \right)}. \quad (3)$$

where I_{max} is the light intensity constant component travelled through the biological object (body part); I_{min} is the minimum intensity, which is caused by a pulse change in the filling of the vessels of the object with arterial blood.

When using (3), light absorption only by the layer of arterial blood is taken into account, which is additionally created during blood filling of vessels. The layer thickness periodically changes from 0 to a maximum value d_{max}. Then, according to (2), the light intensity at the object output using a red LED:

$$I_{min,1} = I_{max,1} \cdot 10^{-d_{max}(\varepsilon_{1,HbO_2} C_{HbO_2} + \varepsilon_{1,HbH} C_{HbH})}. \quad (4)$$

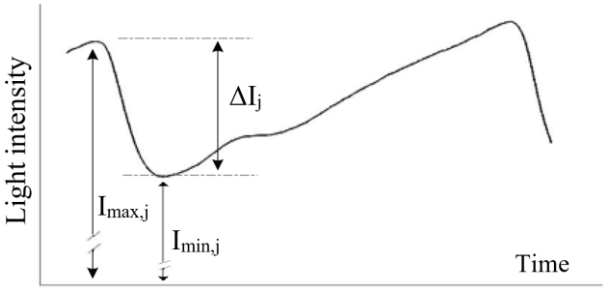

Fig. 1. Time variation of the light intensity travelled through the biological object caused by pulse changes in the filling of vessels of the object with arterial blood [9]

Light intensity at the object output using an infrared LED:

$$I_{min,2} = I_{max,2} \cdot 10^{-d_{max}(\varepsilon_{2,HbO_2} C_{HbO_2} + \varepsilon_{2,HbH} C_{HbH})}. \quad (5)$$

Substituting (4) and (5) into (3), we obtain an expression for calculating the value R:

$$R = \frac{\varepsilon_{1,HbO_2} C_{HbO_2} + \varepsilon_{1,HbH} C_{HbH}}{\varepsilon_{2,HbO_2} C_{HbO_2} + \varepsilon_{2,HbH} C_{HbH}}, \quad (6)$$

and the hemoglobin derivative concentrations interrelation:

$$C_{HbO_2} = C_{HbH} \frac{R\,\varepsilon_{2,HbH} - \varepsilon_{1,HbH}}{\varepsilon_{1,HbO_2} - R\,\varepsilon_{2,HbO_2}}. \quad (7)$$

Substituting (7) into (1), a well-known expression for calculating the functional saturation SpO_2 with oxygen using the auxiliary value R [10]:

$$SpO_2 = \frac{R\,\varepsilon_{2,HbH} - \varepsilon_{1,HbH}}{R(\varepsilon_{2,HbH} - \varepsilon_{2,HbO_2}) + \varepsilon_{1,HbO_2} - \varepsilon_{1,HbH}}. \quad (8)$$

Technically, the task of determining the blood oxygenation level is performed in a PO by measuring the constant and pulse-modulated light components travelled through the tissue in the red and infrared ranges, for further calculation of the R ratio (6) and saturation (8) [11].

However, in addition to functional fractions, human arterial blood also contains dysfunctional fractions that cannot attach oxygen or are already combined with other gases. These include carboxyhemoglobin (HbCO), methemoglobin (HbMet) and sulfhemoglobin (HbSulf).

HbCO is a compound of hemoglobin and carbon monoxide [12]. Excessive HbCO in the blood leads to oxygen starvation, dizziness, nausea, unconsciousness or even death, as carbon monoxide connected to hemoglobin prevents it from attaching oxygen to itself. Normally,

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HbCO concentration in the blood is 0.5-1.5% of total hemoglobin. However, under certain conditions and circumstances, the values are higher. For example, smokers, inhabitants of large cities and highlands, and newborns have HbCO high concentrations (up to 9%) [13]. A separate case in which the HbCO concentration is significantly higher than its normal values is carbon monoxide poisoning [14, 15]. Patients with COVID-19 also have increased HbCO concentrations in arterial blood [16].

HbCO absorbs light when measuring the oxygenation level using the optical method, which causes errors in the measuring results [17]. HbCO high concentrations in arterial blood increase its contribution to the optical density of biological objects, which affects the determination of saturation accuracy only by functional fractions. In addition, changes in HbCO concentration do not affect the value of functional saturation, and this indicator loses its diagnostic value, for example, in the hypoxemia case caused by carbon monoxide poisoning [18].

Therefore, measuring the blood oxygenation level, especially in human body pathological conditions [19], as well as increasing the diagnosis accuracy, is impossible without considering the HbCO concentration in the patient's arterial blood.

1.3. Objective and tasks

The objective is to expand the pulse oximeter functionality to consider the concentration of carboxyhemoglobin in arterial blood.

The research tasks:

1. Development of the theoretical statements for measuring blood oxygenation level considering the HbCO concentration, as well as the structural diagram and mathematical support of the PO for this measurement.

2. Development of the computer model of a well-known PO and a PO with an additional LED that measures blood oxygenation level considering the HbCO concentration.

3. Conduct simulation research using the developed models to determine the effect of HbCO content in arterial blood on the readings of the devices and estimate the device accuracy for measuring the blood oxygenation level considering the HbCO concentration.

2. Materials and methods of the research

2.1. Theory for measuring blood oxygenation level considering the HbCO concentration

Considering the HbCO concentration, oxygen saturation is determined by the following equation:

$$SpO_2(HbCO) = \frac{C_{HbO_2}}{C_{HbO_2} + C_{HbH} + C_{HbCO}}. \quad (9)$$

Equation (9) contains three unknown variables; therefore, determining the saturation requires an additional information channel. The problem is solved using an additional LED with a wavelength that differs from the two main LED light wavelengths.

The auxiliary values R_j are introduced as follows:

$$R_j = \lg \frac{I_{max,j}}{I_{min,j}}, \quad j = 1,2,3. \quad (10)$$

Then, in accordance with the Bouguer-Lambert-Beer law, we have a system of equations:

$$R_j = d_{max} \cdot \sum_{i=1}^{3} \left(\varepsilon_{j,i} \cdot C_i \right), \quad j = 1,2,3. \quad (11)$$

The system of three equations contains 4 unknown variables: the hemoglobin derivatives concentration C_i and the maximum increase in the arterial blood layer per cardiac cycle d_{max}. Therefore, α_i is the product of the concentration of the i-th derivative of hemoglobin by d_{max}:

$$\alpha_i = d_{max} \cdot C_i. \quad (12)$$

Substituting (12) into (11), the system of equations is solved using the Gaussian method relative to α_i. The following:

$$a = \varepsilon_{1,HbH} - \frac{\varepsilon_{1,HbO_2}}{\varepsilon_{2,HbO_2}} \cdot \varepsilon_{2,HbH}; \quad (13)$$

$$b = \varepsilon_{1,HbCO} - \frac{\varepsilon_{1,HbO_2}}{\varepsilon_{2,HbO_2}} \cdot \varepsilon_{2,HbCO}; \quad (14)$$

$$c = \varepsilon_{1,HbH} - \frac{\varepsilon_{1,HbO_2}}{\varepsilon_{3,HbO_2}} \cdot \varepsilon_{3,HbH}; \quad (15)$$

$$d = \varepsilon_{1,HbCO} - \frac{\varepsilon_{1,HbO_2}}{\varepsilon_{3,HbO_2}} \cdot \varepsilon_{3,HbCO}; \quad (16)$$

$$e = R_1 - \frac{\varepsilon_{1,HbO_2}}{\varepsilon_{2,HbO_2}} \cdot R_2; \quad (17)$$

$$f = R_1 - \frac{\varepsilon_{1,HbO_2}}{\varepsilon_{3,HbO_2}} \cdot R_3. \quad (18)$$

$$m = R_1 \cdot (b \cdot c - a \cdot d) - \varepsilon_{1,HbH} \cdot (b \cdot f - e \cdot d) - \varepsilon_{1,HbCO} \cdot (e \cdot c - a \cdot f). \quad (19)$$

Then the values α_i are determined by the following equations:

$$\alpha_{HbO_2} = \frac{m}{\varepsilon_{1,HbO_2} \cdot (b \cdot c - a \cdot d)}; \quad (20)$$

$$\alpha_{HbH} = \frac{b \cdot f - e \cdot d}{b \cdot c - a \cdot d}; \quad (21)$$

$$\alpha_{HbCO} = \frac{e \cdot c - a \cdot f}{b \cdot c - a \cdot d}. \quad (22)$$

To determine saturation, the calculated values α_i are substituted for the concentrations in equation (9):

$$SpO_2(HbCO) = \frac{d_{max}}{d_{max}} \cdot \frac{C_{HbO_2}}{C_{HbO_2} + C_{HbH} + C_{HbCO}} = \frac{\alpha_{HbO_2}}{\alpha_{HbO_2} + \alpha_{HbH} + \alpha_{HbCO}}. \quad (23)$$

Substituting (20) – (22) into (23), a formula for calculating the saturation SpO_2(HbCO), considering the HbCO concentration, is given as follows:

$$SpO_2(HbCO) = \frac{m}{m + \varepsilon_{1,HbO_2} \cdot (f \cdot (b - a) - e(d - c))}. \quad (24)$$

Substituting (20) and (21) into (1), we obtain a formula for calculating the functional saturation SpO_2, considering the HbCO concentration:

$$SpO_2 = \frac{\alpha_{HbO_2}}{\alpha_{HbO_2} + \alpha_{HbH}} = \frac{m}{m + \varepsilon_{1,HbO_2} \cdot (f \cdot b - e \cdot d)}. \quad (25)$$

2.2. Principles of development of a pulse oximeter for measuring the blood oxygenation level considering the HbCO concentration

The block diagram of the simplest PO that measures the blood oxygenation level considering the HbCO concentration is shown in Fig. 2.

The diagram involves the use of three LEDs: red, infrared and additional that are switched on alternately. The LED driver generates signals for alternate switching on of LEDs, which allows the use of one photodetector for light registration. The switching is performed at a frequency of about 2000 Hz. A digital-to-analog converter generates the required LED current value.

The light travels through the biological object (body part), is partially absorbed and recorded by the photodetector. The signal from the photodetector is amplified by an amplifier, converted into a digital signal by an analog-to-digital converter and sent to the microcontroller. The microcontroller processes the signal, calculates the oxygenation level value, displays it, and controls the LED operation via the feedback channel [20].

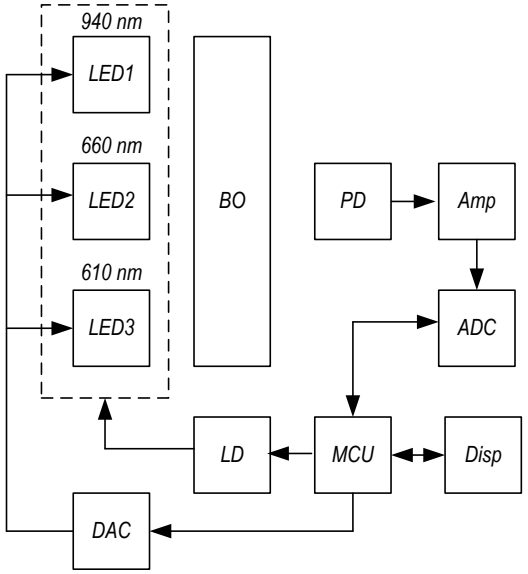

Fig. 2. Block diagram of a PO for measuring blood oxygenation level considering HbCO concentration (BO – biological object (body part), PD – photodetector, Amp – amplifier, ADC – analog-to-digital converter, MCU – Microcontroller, LD – LED driver, DAC – digital-to-analog converter, Disp – display)

The additional LED wavelength choice requires justification. As HbCO extinction coefficient takes smaller numerical values compared with the extinction coefficients of other hemoglobin derivatives [21], it is important that the HbCO contribution to the total optical density of the biological object be the highest at the selected wavelength. Therefore, the following expression is chosen as the objective function:

$$q(\lambda) = \frac{\varepsilon_{\lambda,HbCO} C_{\%,HbCO} \cdot 100\%}{\varepsilon_{\lambda,HbO_2} C_{\%,HbO_2} + \varepsilon_{\lambda,HbH} C_{\%,HbH} + \varepsilon_{\lambda,HbMet} C_{\%,HbMet}}. \quad (26)$$

The function graph (26) is shown in Fig. 3.

The HbCO contribution to the optical density of biological objects is normally not more than 0.4%. Such a small percentage is explained by the low HbCO concentration in arterial blood, and the low extinction coefficient values.

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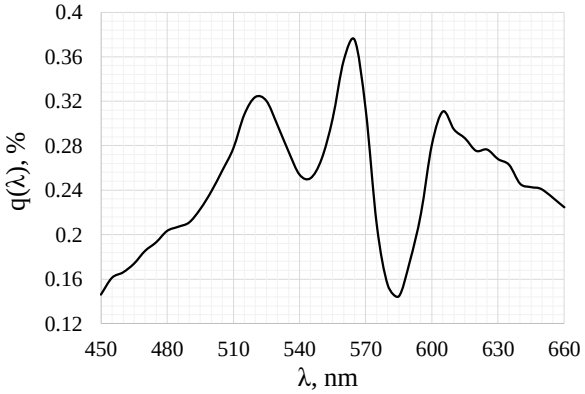

Fig. 3. HbCO contribution dependence of optical density of biological objects on light wavelength

The dependence has three extrema, but two of them are for light with wavelengths of 520 and 565 nm. Such monochromatic light is significantly absorbed by biological objects and cannot be used in POs. Therefore, an LED with a wavelength of 610 nm is chosen as an additional light source (the third extremum of the dependence q(λ)). The light with the chosen wavelength has an amber colour.

To eliminate the influence of outside light on the PO read-outs in the pulse sequence corresponding to each LEDs operation, there is an additional 4th pulse U_4 with the LEDs off. Signal processing in the microcontroller begins with the pulses separation corresponding to the operation of individual LEDs and outside light.

The next set of pulses being received, the calculation of light intensities I_j without components caused by outside light is carried out. To make the calculations, the following formula that considers the photodetector static characteristic nonlinearity is used:

$$I_j = \frac{\lg\left(\left(\frac{U_j}{U_4}\right)^{I_2-I_1}\right)}{\lg\left(\frac{i_2}{i_1}\right)}, \quad j=1,2,3, \qquad (27)$$

where U_j are the separated signals from the photodetector corresponding to a particular LED operation; (I_1; i_1) and (I_2, i_2) are two arbitrary points of the photodetector average nonlinear static characteristic.

The static characteristic of the photodetector in a logarithmic scale is linear. Therefore, in (27) it is necessary to specify the coordinates of only two points of the characteristic.

Extreme values of light intensities I_j were determined. To eliminate local minima and maxima, as well as photodetector signal low-amplitude noise, each determined maximum or minimum value of light intensity is not used in the calculations if the peak height on both sides of the time diagrams is less than a model parameter that is set in advance. In this case, the signal corresponding to the operation of one of the three LEDs is examined for extremes. If there is an extremum on the selected time diagram, there are also extrema on the other two, as the periodic process is caused by the same phenomenon – blood filling of the vessels.

After the next cardiac cycle is completed, using the determined values $I_{max,j}$ and $I_{min,j}$, the auxiliary values R_j are calculated according to formula (10), then the functional saturation and saturation with HbCO are calculated according to formulas (13) – (19), (25), and (24).

To improve measuring accuracy, the blood oxygenation level and heart rate were calculated as average values over several cardiac cycles:

$$SpO_2 = \frac{\sum_{k=1}^{nc} SpO_{2,k}}{nc}, \qquad (28)$$

$$SpO_2(HbCO) = \frac{\sum_{k=1}^{nc} SpO_2(HbCO)_k}{nc}, \qquad (29)$$

$$Pr = \frac{60\cdot(nc-1)}{\sum_{k=2}^{nc}(t_{min,k} - t_{min,k-1})}, \qquad (30)$$

where k is the cardiac cycle serial number, nc is the number of cardiac cycles for calculations, Pr is the heart rate, t_{min} is the time interval between the beginning of the measuring procedure and the achievement of I_{min} in the corresponding cardiac cycle.

2.3. Computer model of the pulse oximeter for measuring the blood oxygenation level considering the HbCO concentration

A computer model of a PO was developed that measures the blood oxygenation level considering the HbCO concentration. The model adequacy is estimated using the interval method [22]. The modelling was carried out using the MATLAB application package and the Simulink graphical modelling environment (Fig. 4).

The PO model with an additional LED consists of subsystems that are models of developed structural diagram individual components (Fig. 2): three LEDs, a biological object with a pulse generator [23, 24], a photodetector, an amplifier, and a microcontroller with the function of displaying the calculated saturation values. A microcontroller model based on «S-Function Builder» block using the C++ language is developed.

Measuring accuracy is estimated by comparing the expected PO read-outs with the initial data, for which the model is completed with blocks for calculating and

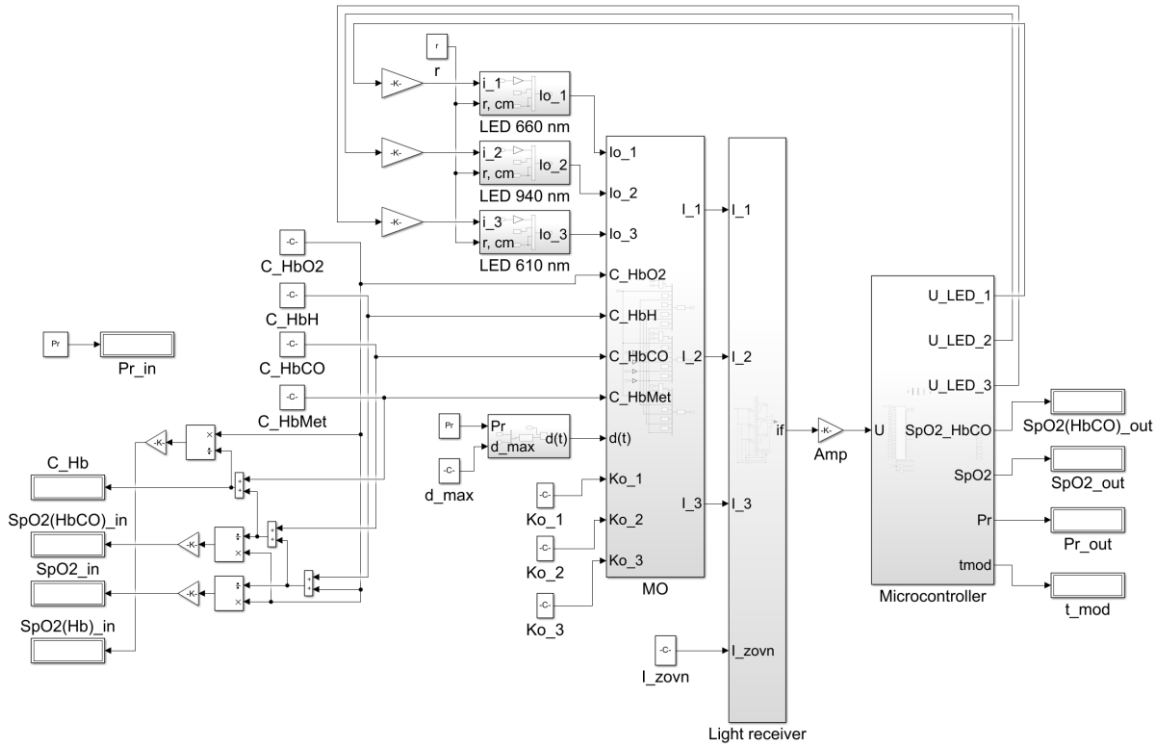

Fig. 4. Computer model diagram of the PO that measures the blood oxygenation level considering for the HbCO concentration

displaying saturation values based on the values of hemoglobin derivative concentrations set during modelling [25].

A computer model of traditional PO was developed to determine the expediency of considering the HbCO concentration when measuring the blood oxygenation level and to estimate the advantages of the proposed device.

There is no third LED subsystem in the model; there is no third control influence from the microcontroller; the pulse number in the sequence is three; there are no extinction coefficients corresponding to the wavelength of the third LED and HbCO in the calculations; the auxiliary value R calculation is carried out according to the known formula (6), and saturation is calculated according to (8).

3. Results

The PO model output values are expected read-outs of functional saturation and saturation with HbCO, heart rate and measuring time.

The input values list, factors of external influence on the measuring process, model parameters, and numerical values that must be set for calculations are given in Table 1. The model considers the main factors affecting the measuring process, namely, the methemoglobin concentration in the arterial blood of patients, the maximum increase in the thickness of the arterial blood layer per cardiac cycle, and changes in the outside light intensity. The symbols and their nominal numerical values are given.

Table 1

Nominal values of the input variables and parameters of the PO model

Variable	Nominal value	Measuring units
Input variables:		
Heart rate, Pr:	60	1/min
Oxyhemoglobin concentration, C_{HbO2}:	96.5 144.75 $8.9834 \cdot 10^{-3}$	% g/l mol/l
Deoxyhemoglobin concentration, C_{HbH}:	2.5 3.75 $2.3273 \cdot 10^{-4}$	% g/l mol/l
HbCO concentration, C_{HbCO}:	0.5 0.75 $4.6546 \cdot 10^{-5}$	% g/l mol/l
Factors of external influence on the measuring process:		
Methemoglobin concentration, C_{HbMet}:	0.5 0.75 $4.6546 \cdot 10^{-5}$	% g/l mol/l
The maximum increase in the thickness of the arterial blood layer in the biological object during one cardiac cycle, d_{max}	0.005	cm
External light intensity, I_{zovn}	0.1	mW/cm^2

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Continuation of Table 1

Variable	Nominal value	Measuring units
Model parameters:		
The LEDs wavelengths, λ_1: λ_2: λ_3:	 660 940 610	nm
PO operating frequency, f	2000	Hz
LED supply voltage, U_{LED}	5	V
Bit depth of the microcontroller ADC, Res	16	bit
The minimum allowable peak height on time diagrams, Tr	2	%
Number of cardiac cycles for measuring, nc	5	
The light extinction coefficients with an emission wavelength λ_1 (660 nm), $\varepsilon_{1,HbO2}$: $\varepsilon_{1,HbH}$: $\varepsilon_{1,HbCO}$: $\varepsilon_{1,HbMet}$:	 319.6 3227 104.41 3706.65	l/(mol·cm)
The light extinction coefficients with an emission wavelength λ_2 (940 nm), $\varepsilon_{2,HbO2}$: $\varepsilon_{2,HbH}$: $\varepsilon_{2,HbCO}$: $\varepsilon_{2,HbMet}$:	 1214 693.39 40 3480	l/(mol·cm)
The light extinction coefficients with an emission wavelength λ_3 (610 nm), $\varepsilon_{3,HbO2}$: $\varepsilon_{3,HbH}$: $\varepsilon_{3,HbCO}$: $\varepsilon_{3,HbMet}$:	 1506 9444 530.86 12766.17	l/(mol·cm)
Total hemoglobin concentration, CHb:	150 $9.3093 \cdot 10^{-3}$	g/l mol/l
The attenuation coefficients by the biological object with emission wavelengths λ_1, λ_2 and λ_3, $K_{o,1}$: $K_{o,2}$: $K_{o,3}$:	 10 10 10	

Continuation of Table 1

Variable	Nominal value	Measuring units
Distance from the LED to the photodetector, r	1	cm
Rated flows of LEDs with emission wavelengths λ_1, λ_2 and λ_3, i_1: i_2: i_3:	 20 20 20	mA
Light intensity of LEDs at rated flows, I_{v1}: I_{v2}: I_{v3}:	 2.4 2.2 2.8	cd
The coordinates of two arbitrary points of the average static characteristic of the photodetector, $I_{f,1}$: $i_{f,1}$: $I_{f,2}$: $i_{f,2}$:	 1 4 1.5 15	 mW/cm^2 mA mW/cm^2 mA
Resistances of the current-to-voltage converter resistors on the operational amplifier, R_1: R_2: R_3:	 1000 1000 2000	 Ohm Ohm Ohm

Considering the existence of other hemoglobin derivatives in arterial blood, it is necessary to determine the fractional saturation which is calculated as the ratio of oxyhemoglobin concentration to the total hemoglobin content in the blood:

$$SpO_2(Hb)_{in,\%} = \frac{C_{HbO_2}}{C_{HbO_2} + C_{HbH} + C_{HbCO} + \sum C_{Hb}} \cdot 100\%, \quad (31)$$

where $\sum C_{Hb}$ is the sum of the other hemoglobin dysfunctional fraction concentrations, including meta-hemoglobin. The index in is the actual values of the calculated values.

Hemoglobin derivatives concentrations expressed as a percentage of the total hemoglobin content are used to calculate saturations:

$$SpO_{2,in,\%} = \frac{C_{HbO_2}}{C_{HbO_2} + C_{HbH}} \cdot 100\% = \\ = \frac{C_{Hb} \cdot C_{HbO_2,\%}}{C_{Hb} \cdot C_{HbO_2,\%} + C_{Hb} \cdot C_{HbH,\%}} \cdot 100\% = \\ = \frac{C_{HbO_2,\%}}{C_{HbO_2,\%} + C_{HbH,\%}} \cdot 100\%. \quad (32)$$

Considering that the sum of all the hemoglobin derivatives percentage concentrations is 100%, the fractional saturation is equal to the oxyhemoglobin percentage concentration in the patient's arterial blood:

$$SpO_2(Hb)_{in,\%} = C_{HbO_2,\%}. \quad (33)$$

According to the developed computer models, the influence of HbCO concentration in arterial blood on the read-outs of the proposed and known POs is researched. The methemoglobin concentration is 0.5%, deoxyhemoglobin is 2.5%, and the change in HbCO concentration was carried out by a corresponding change in the oxyhemoglobin concentration. The modelling results are shown in Table 2.

The dependencies of expected readings $SpO_2(2LED)_{out,\%}$ of the known and $SpO_2(HbCO)_{out,\%}$ of the proposed POs, as well as the actual fractional saturation value $SpO_2(Hb)_{in,\%}$ on the HbCO concentration in the blood are shown in Fig. 5.

The dependencies of expected readings $SpO_2(2LED)_{out,\%}$ of the known and $SpO_2(3LED)_{out,\%}$ of the proposed POs, as well as the actual functional saturation value $SpO_{2,in,\%}$ on the HbCO concentration in the blood are shown in Fig. 6.

4. Discussion

The known PO absolute error caused by the existence of dysfunctional hemoglobin fractions in the blood is as follows:

$$\Delta SpO_2(2LED)_{\%} = SpO_{2,in,\%} - SpO_2(Hb)_{in,\%} = \\ = SpO_{2,in,\%} \cdot (C_{HbCO} + \sum C_{Hb}), \quad (34)$$

and the relative value is equal to the dysfunctional hemoglobin fraction percentage concentrations sum:

$$\delta SpO_2(2LED)_{\%} = \frac{\Delta SpO_2(2LED)_{\%}}{SpO_{2,in,\%}} \cdot 100\% = \\ = C_{HbCO,\%} + \sum C_{Hb,\%}. \quad (35)$$

The proposed PO absolute error caused by the existence of dysfunctional hemoglobin fractions in the blood is as follows:

$$\Delta SpO_2(HbCO)_{\%} = \\ = SpO_2(HbCO)_{in,\%} - SpO_2(Hb)_{in,\%} = \\ = SpO_2(HbCO)_{in,\%} \cdot \sum C_{Hb}, \quad (36)$$

Table 2

The POs operation modelling results at different values of HbCO concentration in the blood

Concentration HbO_2, %	Concentration HbH, %	Concentration HbCO, %	Concentration HbMet, %	Actual value of the functional saturation	Readings of the known PO	Readings of the proposed PO of functional saturation	Actual value of the oxygen saturation considering the HbCO concentration	Readings of the proposed PO of the oxygen saturation considering the HbCO concentration	Actual value of the fractional saturation
67	2.5	30	0.5	96.4	94.63	95.92	67.34	66.38	67
77	2.5	20	0.5	96.86	95.63	96.42	77.39	76.18	77
87	2.5	10	0.5	97.21	96.43	96.8	87.44	85.99	87
89	2.5	8	0.5	97.27	96.56	96.88	89.45	87.85	89
92	2.5	5	0.5	97.35	96.76	96.98	92.46	90.77	92
94	2.5	3	0.5	97.41	96.89	97.03	94.47	92.88	94
95	2.5	2	0.5	97.44	96.95	97.07	95.48	93.69	95
96	2.5	1	0.5	97.46	97.01	97.1	96.48	94.68	96
96.5	2.5	0.5	0.5	97.47	97.04	97.11	96.98	95.19	96.5
96.9	2.5	0.1	0.5	97.48	97.06	97.12	97.39	95.66	96.9
97	2.5	0	0.5	97.49	97.07	97.12	97.49	95.73	97

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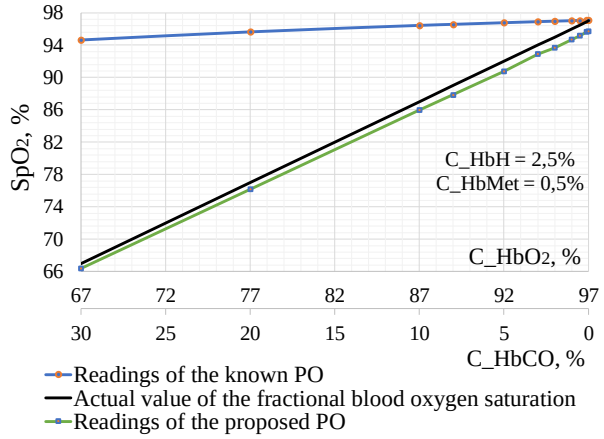

Fig. 5. The dependencies of expected readings of the known and proposed POs, and the fractional saturation actual value on the HbCO concentration in the blood

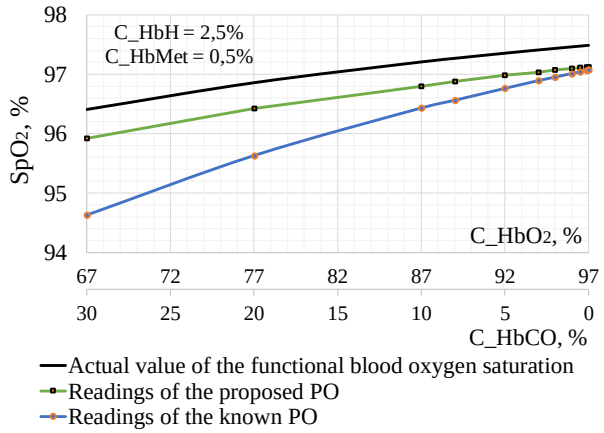

Fig. 6. The dependencies of expected readings of the known and the proposed POs, and the functional saturation actual value on the HbCO concentration in the blood

and the relative error is equal to the dysfunctional hemoglobin fractions percentage concentrations sum excluding HbCO:

$$\delta SpO_2(HbCO)_{\%} = \frac{\Delta SpO_2(HbCO)_{\%}}{SpO_{2,in,\%}} \cdot 100\% = \sum C_{Hb,\%}. \quad (37)$$

Thus, the relative error of the proposed pulse oximeter with an additional LED, unlike the known pulse oximeter error, does not depend on the HbCO concentration in the patient's arterial blood. Taking into account the HbCO concentration allows us to reduce the relative error by $1+\frac{C_{HbCO,\%}}{\sum C_{Hb,\%}}$ times.

The additional LED used and the HbCO concentration considered reduces the absolute measuring error. The difference is

$$\Delta SpO_{2,\%} = SpO_{2,in,\%} - SpO_2(HbCO)_{in,\%} = \frac{C_{HbCO}}{1-\sum C_{Hb,\%}} \cdot SpO_{2,in,\%}. \quad (38)$$

The modelling results (Fig. 5) show that with 10% HbCO concentration and 0.5% normal average methemoglobin value, the well-known PO displays 96% saturation value. That is, the measuring results show that there are no problems with oxygen saturation, whereas only 87% of the total hemoglobin in the blood is combined with oxygen.

The read-outs of the proposed PO with an additional LED that considers the HbCO concentration in the blood, although sensitive to the other hemoglobin dysfunctional fractions concentrations, are close to the actual fractional saturation value. In this case, the expected read-out of such a PO is 86%, which immediately indicates a violation of oxygen transport to body tissues.

In this case, the absolute error of the known PO is 9.43% and the relative error is 10.84%. The absolute error of the proposed PO is 1.01% and the relative error is 1.16%. The errors of the proposed PO are lower than the errors of the known PO by 9 times.

According to the known (8) and proposed (25) dependencies, the same value is calculated – functional saturation, but the calculation results are different (Fig. 6). This is because calculating according to the known formula, there are only two hemoglobin derivatives, oxy- and deoxyhemoglobin in arterial blood, and only these two substances absorb light in an object. The calculations using the proposed formula take into account the additional light absorption by the third derivative. In addition, although the oxyhemoglobin proportion is determined only by functional fractions, the calculation result considers the HbCO concentration.

The absolute error of the known PO in measuring functional saturation is 0.78%, and the relative error is 0.8%. The absolute error of the proposed PO is 0.41% and the relative error is 0.42%. The errors of the proposed PO are lower than the errors of the known PO by 2 times.

Owing to the lack of characteristics or assumptions of carbon monoxide poisoning or HbCO possible increased concentration in the patient blood, it is reasonable to measure the functional saturation. In this case, the use of the proposed PO with an additional LED allows compensating for part of the error that is characteristic of the known PO and caused by HbCO in the blood.

The POs-measuring time is equal to the cardiac cycle duration used to determine saturation and heart rate.

Therefore, the process of measuring the blood oxygenation level, considering the HbCO concentration, does not require an increase in time.

5. Conclusions

1. Technically, the task of measuring the blood oxygenation level considering the HbCO concentration is solved using an additional LED with an emission wavelength of 610 nm in a traditional device. The analytical expressions for calculating oxygen saturation considering the HbCO concentration are determined, which is the basis for the proposed PO mathematical support.

2. Computer models of a traditional PO and a PO with an additional LED are developed, which have the following components: a pulse generator, model of the biological object, LEDs, a photodetector, an amplifier, and a microcontroller. The proposed mathematical support was used to calculate the saturation values in the microcontroller.

3. Based on the developed models, simulation research was conducted to estimate the device accuracy for measuring the blood oxygenation level considering the HbCO concentration. The process of measuring the blood oxygenation level, considering the HbCO concentration, does not require an increase in time. The modelling results show that the developed PO, unlike the known ones, allows detection and estimation of a decrease in blood oxygenation caused by an increase in the HbCO concentration in the patient blood. Considering that light is absorbed not only by two but also by the third derivative of hemoglobin–HbCO – improves the accuracy of the device when measuring functional saturation.

The research materials can be used in the development of technical, mathematical, algorithmic, and software for POs, including POs with additional LEDs that measure the blood oxygenation level, considering the hemoglobin dysfunctional fractions content.

Contributions of authors: conceptualization, methodology – **Anton Sheikus, Igor Prasol, Oleh Datsok**; formulation of tasks, analysis **– Anton Sheikus, Igor Prasol**; development of model, software, verification **– Anton Sheikus, Igor Prasol, Oleh Datsok**; analysis of results, visualization **– Anton Sheikus, Igor Prasol**; writing – original draft preparation – **Anton Sheikus**; writing – review and editing **– Igor Prasol, Oleh Datsok.**

Conflict of interest

The authors declare that they have no conflict of interest in relation to this research, whether financial, personal, authorship or otherwise, that could affect the research and its results presented in this paper.

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Data availability

The manuscript has no associated data.

Use of Artificial Intelligence

The authors confirm that they did not use artificial intelligence methods while creating the presented work.

All the authors have read and agreed to the published version of this manuscript.

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ПУЛЬСОКСИМЕТР ДЛЯ ВИМІРЮВАННЯ РІВНЯ ОКСИГЕНАЦІЇ КРОВІ З ВРАХУВАННЯМ КОНЦЕНТРАЦІЇ КАРБОКСИГЕМОГЛОБІНУ: ПРИНЦИПИ РОЗРОБКИ, КОМП'ЮТЕРНА МОДЕЛЬ ТА ОЦІНКА ТОЧНОСТІ

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Предметом дослідження є математичне, алгоритмічне і технічне забезпечення пульсоксиметра, що вимірює рівень оксигенації крові з врахуванням концентрації карбоксигемоглобіну. Мета роботи – розширення функціональних можливостей пульсоксиметра, яке дозволяє врахувати концентрацію карбоксигемоглобіну в артеріальній крові. В роботі використовуються методи імітаційного комп'ютерного моделювання для побудови моделі і оцінки точності пульсоксиметра, який вимірює рівень оксигенації крові з врахуванням концентрації карбоксигемоглобіну. Розроблено теоретичні положення вимірювання рівня оксигенації крові з врахуванням концентрації карбоксигемоглобіну і структурну схему найпростішого пульсоксиметра, що реалізує дане вимірювання. Запропоновано застосування у відомому пульсоксиметрі додаткового світлодіоду, обґрунтовано вибір довжини хвилі його світла за умовою максимізації внеску карбоксигемоглобіну в оптичну густину об'єкта вимірювання. Розроблено комп'ютерні моделі традиційного пульсоксиметра і пульсоксиметра з додатковим світлодіодом, проведено імітаційні дослідження за розробленими моделями і оцінено точність апарату для вимірювання рівня оксигенації крові з врахуванням концентрації карбоксигемоглобіну. Імітаційні дослідження за розробленими моделями довели, що запропонований пульсоксиметр, на відміну від відомого, дозволяє виявити і оцінити зниження рівня оксигенації крові, яке викликане підвищенням концентрації карбоксигемоглобіну в крові пацієнта. Врахування тієї обставини, що світло поглинається і третьою похідною гемоглобіну – карбоксигемоглобіном, підвищує точність запропонованого пульсоксиметра і при вимірюванні функціональної сатурації.

Ключові слова: пульсоксиметрія; оксигенація; карбоксигемоглобін; розробка; комп'ютерна модель; сатурація; гіпоксемія; точність.

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