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Sensor System Concept for Non-Invasive Blood Diagnosis

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Abstract

The Hemoglobin (Hb) concentration in human blood is an important parameter to evaluate the physiological condition. A hemoglobin test reveals how much hemoglobin can be found in the blood. With this information anemia (a low hemoglobin level) and polycythemia vera (a high hemoglobin level) can be a diagnosed and monitored. It is also possible to observe imminent postoperative bleedings and autologous retransfusions. Currently, invasive methods are used to measure the Hb concentration. For this purpose blood is taken and analyzed. The disadvantage of this method is the delay between the blood collection and its analysis, which does not allow a real-time patient monitoring in critical situations. A non-invasive method allows pain free online patient monitoring with minimum risk of infection and facilitates real time data monitoring allowing immediate clinical reaction to the measured data.

"Keywords: hemoglobin, non-invasive, photoplethysmography"

1. Introduction

In the perioperative area, the period before and after surgery, it is essential to measure diagnostic parameters such as oxygen saturation, hemoglobin (Hb) concentration and pulse[1]. The Hb concentration in human blood is an important parameter to evaluate the physiological condition. With this information anemia (a low hemoglobin level) and polycythemia vera (a high hemoglobin level) can be a diagnosed and monitored. It is also possible to observe imminent postoperative bleedings and autologous retransfusions. Currently, invasive methods are used to measure the Hb concentration. For this purpose blood is taken and analyzed. The disadvantage of this method is the delay between the blood collection and its analysis, which does not permit real-time patient monitoring in critical situations. A non-invasive method allows pain free online patient monitoring with minimum risk of infection and facilitates real time data monitoring allowing immediate clinical reaction to the measured data. The absorption of whole blood in the visible and near infrared range is dominated by the different hemoglobin derivatives and the blood plasma that consists mainly of water.[2] It is well known that pulsatile changes of blood volume in tissue can be observed by measuring the transmission or reflection of light through it. This diagnostic method is called photoplethysmography (PPG) The newly developed optical sensor system uses three wavelengths for the measurement of the hemoglobin concentration, oxygenation and pulse. This non-invasive multi-spectral measurement method is based on radiation of near monochromatic light, emitted by light emitting diodes (LED) in the range of 600*nm* to 1400*nm*, through an area of skin on the finger. The sensor assembled in this investigation is fully integrated into a wearable finger clip and allows full wireless operation through on board miniature wireless enabled microcontroller.

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2. Measurement Method

The new developed sensor system allows a non-invasive continuous measurement of hemoglobin concentration, oxygen saturation and pulse which is based on a multi-spectral measurement method. Thereby the area of skin on the finger is transilluminated by monochromatic light which is emitted by LEDs in the range from 600nm-1400nm. The arteries contain more blood during the systolic phase of the heart than during the diastolic phase, due to an increased diameter of the arteries during the systolic phase. This effect occurs only in arteries but not in veins.[3] For this reason the absorbance of light in tissues with arteries increases during systole because the amount of hemoglobin (absorber) is higher and the light passes through a longer optical path length Δd in the arteries.These intensity changes are the so called PPG-waves.[4] The time varying part allows the differentiation between the absorbance due to venous blood (DC part) and that due to the pulsatile component of the total absorbance (AC part). Figure 1 shows the absorption model for light penetrating tissue to sufficient depth to encounter arterial blood.Upon interaction with the tissue the transmitted light is detected non-invasively by photo diodes. Suitable wavelengths were selected for the analyses of relative hemoglobin concentration change and SpO_2 measurement. During the measurement of hemoglobin the absorption should not be dependent on the oxygen saturation. That means that the measurement is only practicable at so called isobestic points where the extinction coefficients of HHb and HbO_2 are identical. One such point is known to exist around 810nm.[5] According to the assumption that red blood cells are mainly filled with water, the absorption coefficient of blood is similar to a solution consisting HHb, HbO_2 and water (H_2O) and the absorption of HHb and Hb02 is indistinguishable to the absorption of H_2O above 1200nm, it is necessary to select a wavelength value in this region above the diagnostic window.[6] (Figure 2) The Beer-Lambert law (equation 1) describes the reduction of light which is travelling through a homogeneous medium containing an absorbing substance, where Io and I are the incident and transmitted light, $\varepsilon(\lambda)$ is the extinction coefficient of the absorbing substance at a specific wavelength, *c* the concentration of the absorbing substance and *d* the optical path length along the medium.

$$I = I_0 e^{-\varepsilon(\lambda)\mathrm{cd}} \quad (1)$$

The Beer-Lambert's law is also valid if more than one absorbing substance is present. Each absorber contributes one part of the total absorbance. The unscattered absorbance A is defined as the negative natural logarithm of the transmittance (ratio of *I* and I_0) of light. The resulting total absorbance A_t of a medium with n absorbing substances yields to equation 2:

$$A_t = \sum_{i=1}^{n} e_i(\lambda)\mathrm{c_i d_i} \quad (2)$$

Therefore, Beer-Lamberts law allows the concentrations of *n* different substances to be determined if the absorbance of light is measured at n different wavelengths and the extinction coefficient of the substances are known. [7] The measuring principle of pulse oxymetry is based on the different absorbance properties of oxygenated and desoxygenated hemoglobin and the pulsatile intensity variation. The volume and pressure fluctuations are generated in the systolic phase of the heart. In the case of pulse oximetry the oxygen saturation is defined as [8]:

$$SO_{2,funct} = \frac{c_{cHbO_2}}{c_{cHbO_2} + c_{HHb}} \quad (3)$$

The light emitting diodes (LEDs) produce different output light intensities. For comparison of the two wavelengths a normalisation is therefore necessary. As a result of the normalisation the optical path length (if d λ_1 = dλ_2) through the tissue and the incident intensity I_0 are no longer significant. The ratio of the time varying to non-time varying (AC/DC) values of the both wavelengths used results in the ratio R:

$$R = \frac{A_{t,RED}}{A_{t,IR}} = \frac{\frac{\mathrm{I_{AC+DC,RED}}}{\mathrm{I_{DC,RED}}}}{\frac{\mathrm{I_{AC+DC,IR}}}{\mathrm{I_{DC,IR}}}} = \frac{\varepsilon_{\mathrm{HHb}}(\lambda_{RED}) + \mathrm{SaO_2}[\varepsilon_{\mathrm{HbO_2}}(\lambda_{RED}) - \varepsilon_{\mathrm{HHb}}(\lambda_{RED})]}{\varepsilon_{\mathrm{HHb}}(\lambda_{IR}) + \mathrm{SaO_2}[\varepsilon_{\mathrm{HbO_2}}(\lambda_{IR}) - \varepsilon_{\mathrm{HHb}}(\lambda_{IR})]} \quad (4)$$

This theoretical equation leads to incorrect results in practice, because the optical properties given of human blood are much more complicated. For this reason all pulse oxymeters must be calibrated. One possibility is the replacement of the extinction coefficients with constants. These constants are determined and validated in clinical studies, mostly using invasive measurement method. Based on the pulse oxymetry derivation the haemoglobin determination will be performed at the wavelengths λ_1 = 810nm and λ_2 = 1300nm.[9] The AC/DC Values of both wavelengths leads to the quotient H (equation 5):

$$H = \frac{\frac{I_{AC+DC,810nm}}{I_{DC,810nm}}}{\frac{I_{AC+DC,1300nm}}{I_{DC,1300nm}}} = \ln(10)\frac{\varepsilon_{Hb}(\lambda_{810nm})c_{Hb}}{\mu_{H20}(\lambda_{1300nm})64500\mathrm{g/mol}} \quad (5)$$

This theoretical equation results in incorrect values in practice but the fact f(H) is true[10]. Thus, the hemoglobin sensor needs calibration as in the case of the pulse oxymeter.

3. Sensor System

The sensor system being developed consists of hardware modules including appropriate light sources and receivers, a microcontroller and a wireless interface. A key component allowing low power wireless operation is the low power microcontroller MSP430.

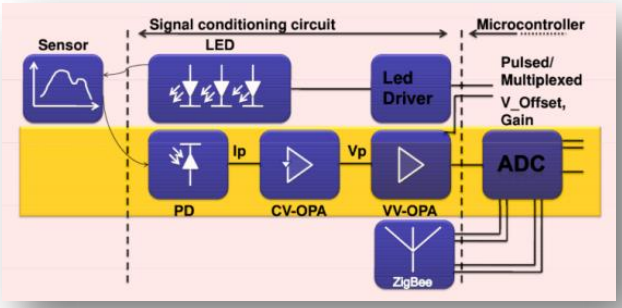

Figure 1: Functional Diagram Sensor Device Figure 2: Prototype Hb-Sensor System

This enables software controlled and time multiplexed operation of the light sources and receiver channels, multiplexed operation of the light sources and receiver channels. The mean value is calculated and dark current subtracted in software, and the data is transmitted via USB or wireless interface. With application software programmed in LabVIEW it is possible to handle the data on a Laptop or PC. The light sources, three LEDs, with centre wavelengths of λ_1= 670nm, λ_2= 810nm and λ_3= 1300nm are installed in the upper part of the clip. The pulsed LED currents are controlled by the microcontroller, which allows a change of the light source intensity. To detect the transmission signals an Indium Gallium Arsenide/Indium-Phosphor photodiode was chosen. With a spectral range of 400nm-1700nm a measurement at all three used wavelengths is feasible [11]. All components of the sensor system are chosen to guarantee low power consumption. Figure 2 shows a picture of the sensor system.

4. Results and Conclusion

First measurements with the new hemoglobin sensor system of the transmission signals on the finger terminal element have shown variations in light absorption due to the arterial pulse at all three wavelengths. [12] The relative

attenuation coefficients R and H were offline calculated and first results are showing that the newly developed sensor device is able to measure the SpO_2 and hemoglobin concentration. Figure 3 shows the signal at 1300nm, 810nm and 670nm. Based on human circulatory system a bloodstream model has been designed which is necessary for validation of the measurement method of hemoglobin concentration and the newly developed optical sensor system. With the help of the model a controlled variation of the blood parameter, hemoglobin concentration and oxygen saturation are feasible. Thereby both parameters can be changed separately or simultaneously.
The developed sensor device is suitable for non-invasive continuous online monitoring. The advantage of this measuring technique is independent of blood samples and this allows a minimum risk of infection and it makes possible to react immediately on the measured data.

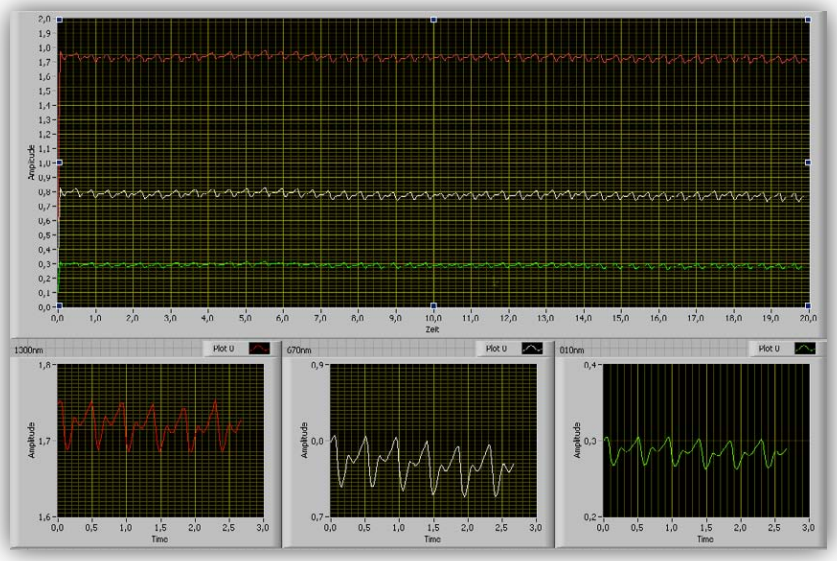

Figure 3: PPG raw signals at 1300nm, 810nm and 670nm

Acknowledgements

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