Remote Optical Monitor of Blood Oxygenation

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***Abstract* - Pulse oximetry is a widely used technique in biomedical analyses. It enables monitoring the percentage of saturated hemoglobin and heart pulse in a noninvasive way. The mandatory permanence of patients in medical facilities due to continuous measurements of blood oxygen saturation and cardiac pulse can be avoided by taking advantage of nowadays telecommunications technologies. This paper describes a pulse oximeter capable of operating in standalone mode and wirelessly transmit the data to a display unit.**

***Index Terms*— Pulse oximeter, optical sensors, Bluetooth™, Java.**

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I. INTRODUCTION

Pulse oximetry has become the standard technique for monitoring oxygenation during procedural sedation-anesthesia, therapy in the intensive and neonatal care unit, and recovery from anesthesia in the postanesthesia care unit. Oximetry is the spectrophotometric determination of the hemoglobin-oxygen saturation [1]. A pulse oximeter monitors, calculates, and displays the measured hemoglobin oxygen saturation as a percentage - SpO2 - and the cardiac pulse. This technology has been rapidly adopted throughout the health care community due to its easy use and importance of the information provided continuously and noninvasively allowing real-time monitoring.II. Another heading (The 1ª letter is caps)

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II. PULSE OXIMETRY

A. History

In 1929, the American physiologist, Glen Millikan, studying at Cambridge, optically measured the speed of O_2 combination, using purple and yellow filters. In 1935, Karl Matthes, the son of a hospital administrator and a professor of physiology from Vienna, using light with two wavelengths, constructed the first device to continuously measure the SpO_2 of human blood in vivo, and stated its principle that ‘‘Red light can pass through oxy-hemoglobin but reduced hemoglobin absorbs it’’ [2]. The first modern pulse oximeter was developed, in 1972, by Takuo Aoyagi, a Japanese bioengineer that created the idea of measuring only the pulsatile changes in light transmission through living tissues to compute the arterial O_2 saturation. It was this key idea that permitted the development of instrumentation that required no calibration after its initial factory setting, as all human blood has essentially identical optical characteristics in the red and infrared bands used in pulse oximetry [3].

B. Principles

The pulse oximeter uses light emitting diodes (LED emitters) to generate optical signals in two wavelengths — one at 660 nm and the other at 880 nm (Figure 1 displays the emission spectra of the two LED emitters).

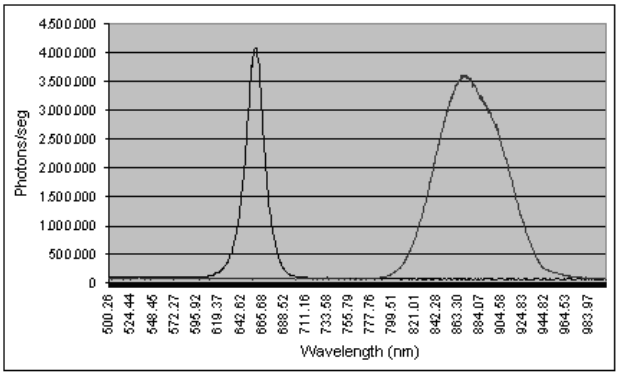

Fig. 1 Light emitting diodes spectrum of the used oximeter probe.

Hemoglobin and other tissues such as bone, fat, skin, and fingernails between the LED emitters and the detector absorbs these red and infrared light wavelengths. The amount of light absorbed varies, depending on the pulsating arterial flow in the tissues and the saturation level of the hemoglobin in the erythrocytes (Figure 2). The pulse oximeter calculates the absorption based on two physical principles: the two wavelengths of light are absorbed differently by oxygenated hemoglobin and deoxygenated hemoglobin (Figure 3); and tissues such as bone, fat and venous blood absorb a relatively constant amount of light, producing a relatively constant rate of light absorption. The ratio of absorption at these two wavelengths varies with the oxygen saturation. This ratio is converted to a numerical SpO_2 value through the use of calibration curves derived from volunteer desaturation studies [4]. All pulse oximeters require a pulsatile signal at the sensor site to generate a digital value on the display.

The pulsatile characteristic of the signal is produced by the fluctuating volume of arterial blood between the emitter and detector in the sensor. In situations involving decreased blood flow and/or high peripheral vascular resistance, the pulse oximeter may be unable to display an accurate measurement.

The probe structure includes both LED emitters and the photodetector that will receive the transmitted light. The probe acquires both pulsatile signals from the patient and extraneous artifacts from a variety of sources. Both are forms of energy. Any external energy captured by the sensor may produce artifact.

Some pulse oximeters use reflected light instead of transmitted, as in our case. The only advantage of those pulse oximeters is the fact that the probe can be positioned virtually everywhere in the human body.

The pulse oximeter developed in this project was a transmission pulse oximeter.

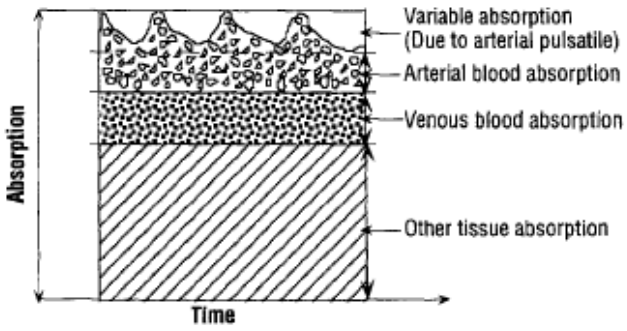

Fig. 2 Partial contributions in total absorption [4].

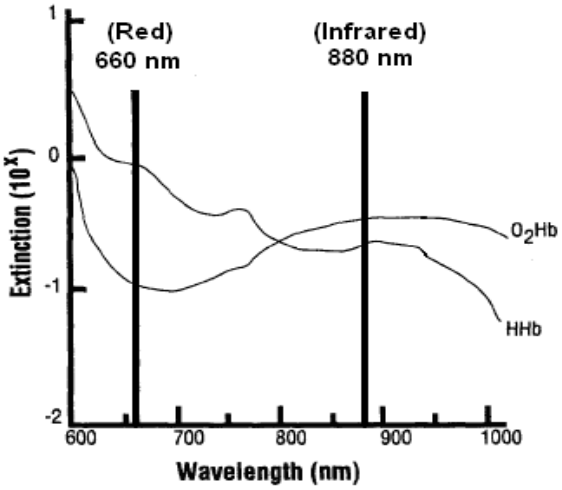

Fig. 3 Absorption of oxygenated hemoglobin and deoxygenated hemoglobin at Red and Infrared wavelength [4].

Mathematical models

Nowadays, pulse oximeters rely on empirical calibration approaches, which are very accurate at saturations above approximately 70 % but perform poorly at low saturations [5].

The percentage oxygen saturation is calculated as a function of the transmitted light normalized time varying components ratio.

$$R = \left[\frac{\frac{AC_{RED}}{DC_{RED}}}{\frac{AC_{IR}}{DC_{IR}}} \right] \quad (1)$$

where AC_{RED} and DC_{RED} are the ac and dc levels of the red led, and AC_{IR} and DC_{IR} are the ac and dc levels of the infrared led [5].

The Beer-Lambert model combines August Beer (1825-1863) law and Johann Lambert (1728-1777) third law of propagation of light in a transparent matter [6].

Using the principle of Beer-Lambert's law, the concentration of a given solute in a solvent is determined by the amount of light that is absorbed by the solute at a specific wavelength [2]

$$I = I_0 \times 10^{sCD} \quad (2)$$

where I_o *is* the incident light intensity, I is the transmitted light intensity, C is the medium concentration, D is the depth and ε is the extinction coefficient. However the Beer-Lambert model has been overrun by the more accurate photon diffusion models. The need of a reliable mathematical models for saturations below 70 % has lead to the use of photon diffusion theory in pulse oximetry.

The transmission of light trough biological tissue is dependent on many factors, including thickness, tissue type, blood content and other materials. Therefore, the total attenuation of light trough tissue is due to the absorption from chromophores in the tissue, as well as light scatter. The Beer-Lambert states that, in the absence of scattering conditions, the total attenuation is a linear sum of that due to each chromophore. However, biological tissues are highly scattering and thus, wavelength dependent sources of attenuation, such as path length and absorption are introduced [7]. Are those wavelength dependent absorption properties of the various natural chromophores that make possible the quantification of blood and tissues oxygenation by optical spectroscopy [8].

Although the photon diffusion theory is more complex it models the response of a transmission pulse oximeter more accurately than the Beer-Lambert model [9].

$$S_pO_2 = \frac{R\sigma_{IR}^{Hb} - N\sigma_r^{Hb}}{N\left(\sigma_r^{HbO_2} - \sigma_r^{Hb}\right) + R\left(\sigma_{IR}^{Hb} - \sigma_{IR}^{HbO_2}\right)} \quad (3)$$

where *R is* the transmitted light normalized time varying components ratio, *N* is a function of the attenuation coefficient of the blood-perfused tissues and of finger thickness, σ^{Hb} is the extinction coefficient of

hemoglobin and σ^{HbO_2} is the extinction coefficient of oxygenated hemoglobin.

The response curve for the red/infrared ratio vs. the oxygen saturation is generated by calculating the intensity of transmitted light over a range of oxygen saturations (Figure 4)

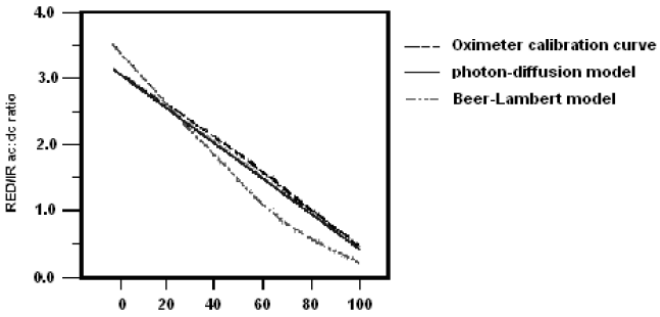

Fig. 4 Red/infrared ratio vs. oxygen saturation for the three different models [10].

The curve used in this project was the empirical calibration approach since the values of saturation expected to be measured were above 70% and for that range of S_pO_2 this model is still the most accurate.

III. SYSTEM IMPLEMENTATION

The pulse oximeter consists of a sensor, a control unit for the sensor which calculates certain parameters (such has the SpO_2) and a display (Figure 5).

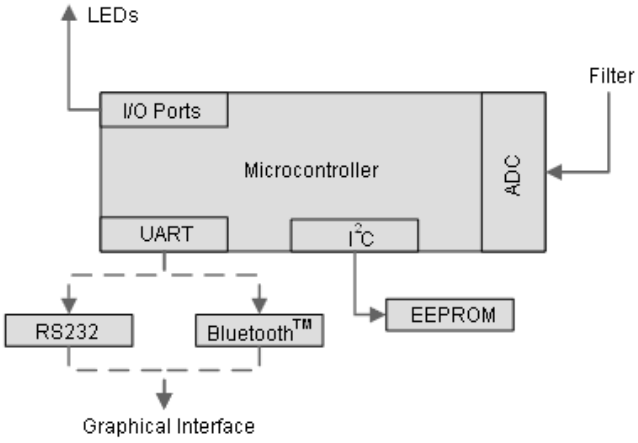

Fig. 5 Simplified hardware architecture.

The LEDs are time multiplexed for distinction of results, Figure 6. The remaining time of the period is used by the microcontroller to perform signal processing.

The microcontroller controls and samples the LED emitters output. Since the interval for processing is limited by the remaining time of the 1 ms period (about 800 ms), the microcontroller does not operate in polling mode, but in interruption mode. It also processes the data, saves it in

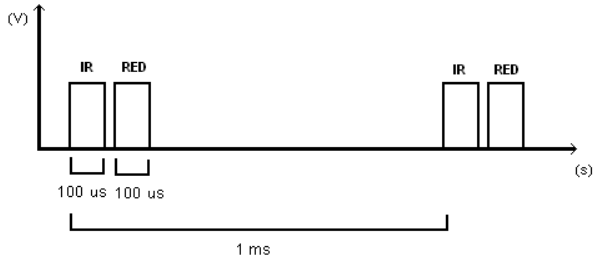

Fig. 6 LEDs operation

an Electrically Erasable Programmable Read Only Memory) (EEPROM), if necessary, and manages the communication between the module and the display unit.

To bias correctly the LEDs for each person, the microcontroller is combined with a quadruple switch to provide four different current for each LED.

The received signal is converted to voltage by a current-to-voltage converter, then is filtered by a simple Butterworth lowpass filter. This signal is the input of the analog-to-digital converter (ADC). After being processed, S_pO_2, heart rate and instant of measurement is saved in an external EEPROM if no connection is established with a display device, making data available for later use. The communication with the EEPROM is made using I^2C^{TM} protocol. It is a serial protocol developed by Phillips Semiconductors in 1980 and uses 2 wires, one for synchronism between devices and the other for data transfer.

To improve the ADC resolution, the reference voltages of conversion (Vref+ and Vref-), had 1 Volt difference from each other. Therefore, the 10 bit ADC will convert a span of 1V into 1024 digital values, instead of the available 3.3 V. To perform that, we have emulated a Pulse Witdh Modulation (PWM) that is correlated with the received signals from the LEDs. This PWM is combined with two second order Butterworth lowpass filters to produce Vref+ and Vref-. Limiting the distance of amplitude between the received signals from the LEDs to 800 mV, we guarantee that these signals will always fit between Vref+ and Vref- (Figure 7).

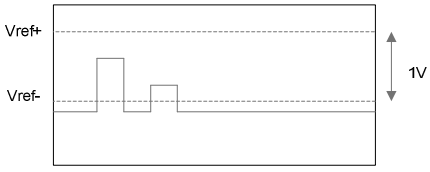

Fig. 7 ADC range of conversion

There are two possible ways of transmitting data to the display device. Both are series communication but the transmission medium is different. RS232 uses, in our case, a 4 wire cable and Bluetooth™, as a wireless technology, uses air.

The final setup (Figure 8) has a length of 6.3cm, a width of 4.2 cm and 1.6 cm of high when the Buetooth™ module is included. The setup is powered at 3.3 V by a small mobile battery, which ensures mobility and autonomy.

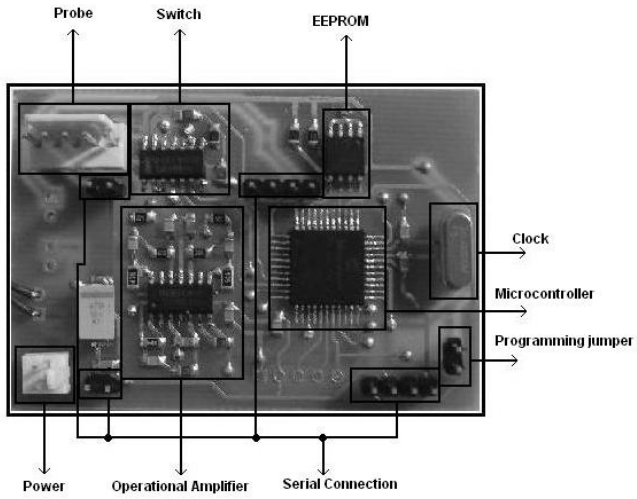

Fig. 8 Hardware implementation.

The display unit shows the parameters calculated by the microcontroller. These parameters include the S_pO_2, the heart rate and the red and infrared data obtained from the sensor - the plethysmographic wave (Figure 9).

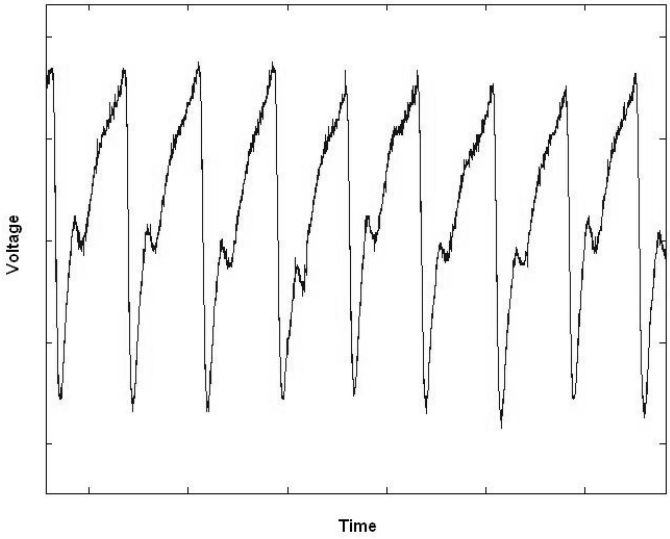

Fig. 9 Plethysmographic wave obtained from RED LED.

The display unit may be a Personal Computer (PC), a laptop or a mobile device, such has a cellular phone or a Personal Digital Assistant (PDA).

Two graphical applications were created: one for PCs and another for mobile devices.

For each application, the display has included the option to define alarms so that the user can be alerted by the display unit of special occurrences.

The communication between the display device and the oximeter can be established using a simple RS232 serial link or a Bluetooth™ wireless. The Bluetooth™ link was exhaustively tested to prevent errors in the transmission of data.

To guarantee the maximum compatibility, the display application was programmed in Java language for both applications (Figure 10). But it was not possible, at this moment, to visualize any graphics in the mobile application however quantitative online data was displayed in text (Figure 11).

The microcontroller/probe module can also work not connected to the display unit. In this mode, it can gather information for up to approximately 16 hours in the EEPROM. The data gathered can be visualized by any display unit at any time. This unit has the option to generate an event each time the predefined alarm occurs. Consequently, it is not necessary to analyze the full 16 hours of recorded data.

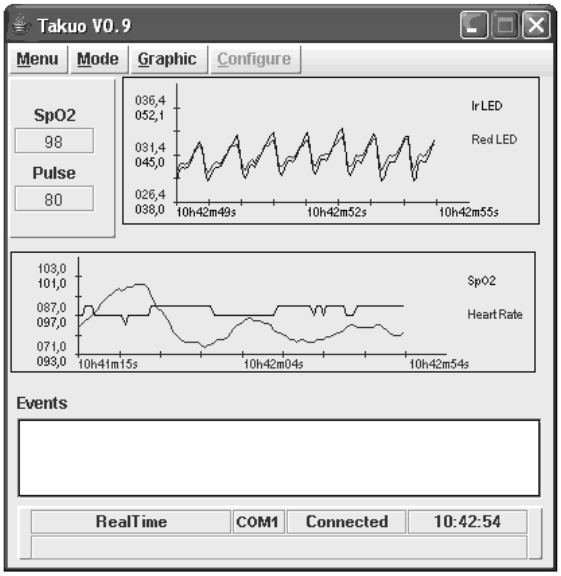

Fig. 10 Layout of the Java interface: at the top right the plethysmographic wave, at the center a graph with the last 90 second of blood oxygen saturation and cardiac pulse and at the bottom a text box for signaling the alarm events.

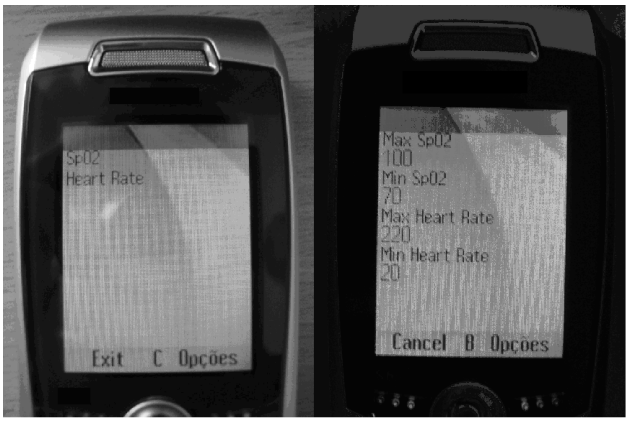

Fig. 11 Layout of mobile interface

IV. CONCLUSION

The studied oximeter was compared with commercial units, which are found in medical facilities, and the values of blood oxygen saturation and cardiac pulse were identical. In some cases, a slight variation up to 2 % was

noticed. These differences could be originated by large movement artifacts, displacement of the probe or even signal corruption due to environment light.

The wireless communication via Bluetooth™ proves to be a reliable alternative to split the microcontroller module, which performs the acquisition and processing of all data, from the display device. However, the transmission rate of RS232 is higher then the Bluetooth™ module tested. Therefore, in the RS232 case, additional data can be sent to the graphical application to design a smoother plethysmographic wave.

The choice of Java language permits the use of the display application on a wide range of devices since most of present day devices, such as, Personal Computers, laptops, cellular phones and Personal Digital Assistants support Java Virtual Machine that is required for Java applications.

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REFERENCES

[1] D. Hatlestad, "Revolution in Pulse Oximetry Enhances Respiratory Patient Care. RT The Journal for Respiratory Care Practitioners®" October/November 2001 Available: http://www.rtmagazine.com.

[2] V. Kamat, "Pulse Oximetry", Indian Journal Anesthesia, 2002.

[3] Y. Pole, "Evolution of the pulse oximeter", International Congress Series, pp. 137-144, 2002.

[4] J. E. Sinex, "Pulse oximetry: Principles and Limitations", American journal of medicine, Vol. 1, January 1999.

[5] G. Zonios, U. Shankar and V. K. Iyer, "Pulse Oximetry Theory and Calibration for Low Saturations", IEEE Transactions on Biomedical Engineering, Vol. 51, Nº 5, May 2004.

[6] D. Kerr, H. Hhalil, L. McGill, "Noninvasive pulse oximetry: physiology and engineering", October 2003.

[7] T. Mawn, P. Li, S. Nioka, and B. Chance, "Trans-abdominal monitoring fetal arterial oxygen saturation using pulse oximetry", IEEE EMBS – NEBE 2002.

[8] S. M. L. Silva, "Application of Near Infrared Spectroscopy to Pulse Oximetry and Tumour Oximetry", Lecture Notes of the ICB Seminars, Vol. 61, 2003.

[9] D. R. Marble and P. W. Cheung, "Mathematical model of transmission pulse oximetry", IEEE Engineering in medicine & biology society 10th annual international conference, 1988.

[10] D. Kerr, H. Hhail and L. McGill, "Noninvasive pulse oximetry: