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A CMOS Camera-Based Pulse Oximetry Imaging System

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Abstract—**In this paper a CMOS camera-based system for non-contact pulse oximetry imaging in transmission mode is described. Attention is drawn to the current uses of conventional pulse oximetry and the potential application of pulse oximetry imaging to developing objective wound assessment systems.**

I. Introduction

Pulse oximetry [1, 2] is the non-invasive measuring of arterial oxygen saturation (SaO_2) using at least two wavelengths of light to determine the relative concentrations of oxyhaemoglobin (HbO_2) and deoxyhaemoglobin (Hb) in the blood. A pulse oximeter probe typically consists of two light emitting diodes and a photo diode detector. The probe can operate by reflection or more commonly by transmission and is usually attached to an extremity (typically a finger or earlobe) by a plastic clamp that serves to maintain good contact with the tissue and shield the detector from extraneous light sources. It plays an important role in monitoring the health of patients during recovery and perhaps most importantly during anaesthesia, where it can provide an early indication of cyanosis [1]. Indeed it has been described as "arguably the most significant technological advance ever made in monitoring the well-being of patients during anaesthesia", in [3]. In addition to its suitability to monitoring gross oxygen deficiency in the body, pulse oximetry is also applicable to monitoring the health and viability of smaller areas of tissue. Pulse oximetry is a sensitive indicator of perfusion and has been shown to be capable of detecting peripheral vascular disease [4]. It has also been used successfully to assess the viability of tissue during surgery and experimentally to discern between ischaemia from which tissue will recover and ischaemia from which tissue necrosis will result [5]. The value of the pulse oximeter in each of these applications is that it provides an objective measure of oxygen saturation. Without the pulse oximeter, judging the level of cyanosis or ischaemia is entirely subjective and depends on the experience of the clinician, the condition of the patient and even the ambient lighting [6].

Another area that stands to benefit from pulse oximetry is wound assessment. Specifically the assessment of wounds such as pressure sores and ulcers that result from ischaemia and poor circulation, the condition of which is currently evaluated using subjective scales. The ability to monitor the oxygen saturation and perfusion of areas likely to succumb to pressure sores could aid in identifying such wounds at an early stage before they are clinically visible. Once they are established these wounds are difficult and costly to treat.

In this paper a CMOS camera-based system is presented that is capable of monitoring perfusion and arterial oxygen saturation changes in an extremity without making contact with the tissue under investigation. The principles of extracting the haemodynamic information from such as system are also described.

II. Methodology

A. Hardware

The transmission mode pulse oximetry imaging system is depicted in Fig. 1. A subject's finger is illuminated on one side by two near infrared (NIR) light-emitting diodes producing light centered at wavelengths of 760 nm and 880 nm. A Pixelink monochromatic PL-A741 2/3" CMOS FireWire camera is located 30 – 50 cm above the finger on the opposite side, focused by a zoom lens (C-Mount, 18 mm – 108 mm, f2.5 – closed). The two diodes are alternately energized (one on, the other off and vice versa). The camera receives a trigger signal when either diode is energized. In this way the transmitted light intensity from each diode is captured in alternate frames. The transmitted light intensity varies with the pulsing of the blood. A plot this variation against time is referred to a photoplethysmograph (PPG). A frame rate of 60 frames per second (30 frames per wavelength per second) provides an appropriate tradeoff between sampling often enough to capture the fine structure of the PPG signals and allowing the camera sufficient time to integrate and read out each frame.

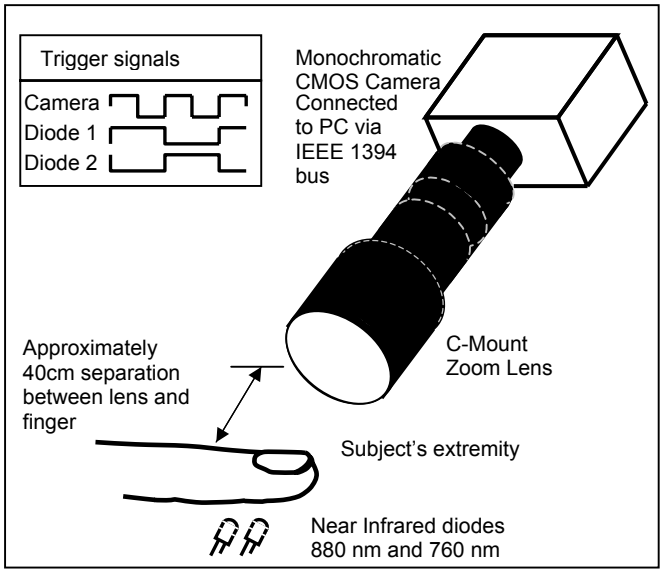

Fig. 1. Transmission mode pulse oximetry imaging system.

When the camera has received a preset number of trigger signals it writes the captured frames to a PC as an uncompressed AVI file.

B. Signal Acquisition

The resulting AVI file is processed using Matlab®. Frames 1,3,5,7… have been captured at 760 nm and frames 2,4,6,8… at 880 nm. The interleaved frames are separated. Each frame is divided into boxes or groups of adjacent pixels. For each frame the average pixel value of each box is calculated. Plotting the average pixel value of each box for each frame yields a PPG signal. This process is illustrated in Fig. 2. The process is performed on both the 760 nm and 880 nm frame sets. The variations in the pixel values in each frame are influenced by both the changes in absorption by the finger as the blood pulses through it and change in the ambient light, to which the camera is also sensitive. Although the camera is more susceptible to interference from ambient light than a conventional contact probe, the PPG signals captured by the camera have been shown to be comparable to those captured by conventional probe [7]. In Fig. 2 (B) the pulsing effect is obvious. During the cardiac cycle at times of high pressure (systole) absorption is at its highest and the average pixel value decreases, during low pressure (diastole) absorption is at its lowest and the average pixel value increases. (Note the waveform has been inverted so it appears as the familiar arterial pressure waveform.) The inflection present on the falling edge of the peaks in Fig. 2 (B) is the dichrotic notch, a short back flow of blood caused by the abrupt closure of the aortic valve. In clinical pulse oximiters the appearance of the PPG waveform is used as an indicator that the probe is properly attached, it is common for such devices to cease displaying an oxygen saturation value if the PPG signal cannot be detected [1].

C. Extracting the Haemodynamics

Weather the data has been acquired through a conventional pulse oximeter probe or by a camera system; the technique used to extract the changes in Hb and HbO_2 is the same. The technique is based on the Modified Beer-Lambert Law. The algorithm used here is based on that described in [8]. The Modified Beer-Lambert Law is described by (1).

$$A = \log_{10}(I_0 / I) = \alpha.c.d.B + G \qquad (1)$$

This law sates that the attenuation (A) measured at the detector as the logarithm of the incident light (I_0) divided by the detected light (I), is equal to the specific extinction coefficient of the absorbing substance (α), multiplied by the concentration (c) of that substance, multiplied by the distance between source and detector (d), multiplied by a differential path length factor (B). The B factor accounts for the fact that due to scattering a photon travels a much greater distance than the geometrical distance between source and detector.

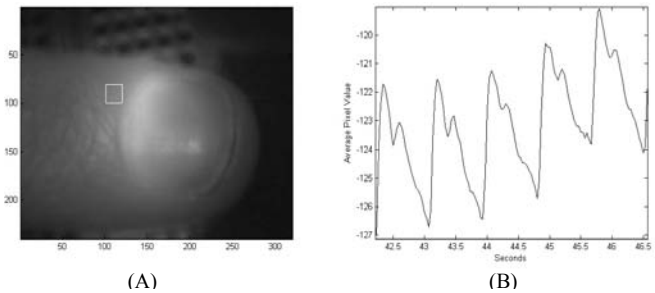

Fig. 2. (A) A sample frame with a 20x20 pixel box outlined. (B) The change in the average pixel value of that box over time.

An additional term is added to the expression to account for loss of photons due to scattering (G). The scattering loss is assumed to be constant for a given tissue. The effective path length ($d.B$) varies little and α is fixed for a specific absorber at any particular wavelength. Thus (assuming I_0 remains constant) by differentiating (1) with respect to time the equation reduces to that described in (2).

$$\Delta A = A_1 - A_2 = \log_{10}(I_2 - I_1) = \alpha.\Delta c.d.B \qquad (2)$$

The values of α, d and B are known for the absorbers and tissues in question, so by measuring the change in light attenuation at a particular wavelength, the change in concentration of the absorber can be determined. Since blood has two significant absorbers (Hb and HbO_2) the change in attenuation of light of wavelength 760 nm over time can be described by (3).

$$\Delta A_{760nm} /(B.d) = \alpha_{760nm,Hb}\Delta c_{Hb} + \alpha_{760nm,HbO_2}\Delta c_{HbO_2} \qquad (3)$$

Similarly the change at 880 nm is given by (4).

$$\Delta A_{880nm} /(B.d) = \alpha_{880nm,Hb}\Delta c_{Hb} + \alpha_{880nm,HbO_2}\Delta c_{HbO_2} \qquad (4)$$

This can be expressed in matrix form as $A/(Bd)=\alpha C$ and solved for C by finding the pseudo-inverse of α, such that $C=(\alpha^T\alpha)^{-1}\alpha A/(Bd)$ (where α^T is the transpose of α).

In the camera based system the distance between the source and detector (d) is taken to be 1.2 cm. Although the actual distance between the diodes and camera is far greater, for most of the distance the photons are traveling through air. The differential path length factor (B) has been shown to be wavelength and age dependent [8]. The differential path length factor for 780 nm is given by $B_{780}=5.13+0.07(\text{age in years})^{0.81}$. To obtain the value of B_{760} and B_{880}, the calculated value of B_{780} is scaled by a factor of 1.12 and 0.84 respective [8].

III. RESULTS

Fig. 3 depicts typical results obtained using the camera pulse oximeter and the haemodynamic analysis algorithm described above.

For a healthy person at normal atmospheric pressure arterial oxygen saturation (SaO_2) is typically around 96-98% [9]. This means that most of haemoglobin molecules in the arteries have oxygen molecules attached (up to four oxygen molecules per one haemoglobin molecule). As a result the change in concentration of HbO_2 over one cardiac cycle is generally much larger than the accompanying change in the Hb concentration. This can be seen in the bottom plot in Fig. 3. Arterial oxygen saturation is usually defined as the amount of haemoglobin sites with oxygen molecules attached divided by the total number of haemoglobin sites available for oxygenation transportation as in (5).

$$SaO_2(\%) = \frac{HbO_2}{HbO_2 + Hb} \times 100 \qquad (5)$$

A percentage reading of SaO_2 can be calculated from the data presented in Fig. 3 by comparing the corresponding

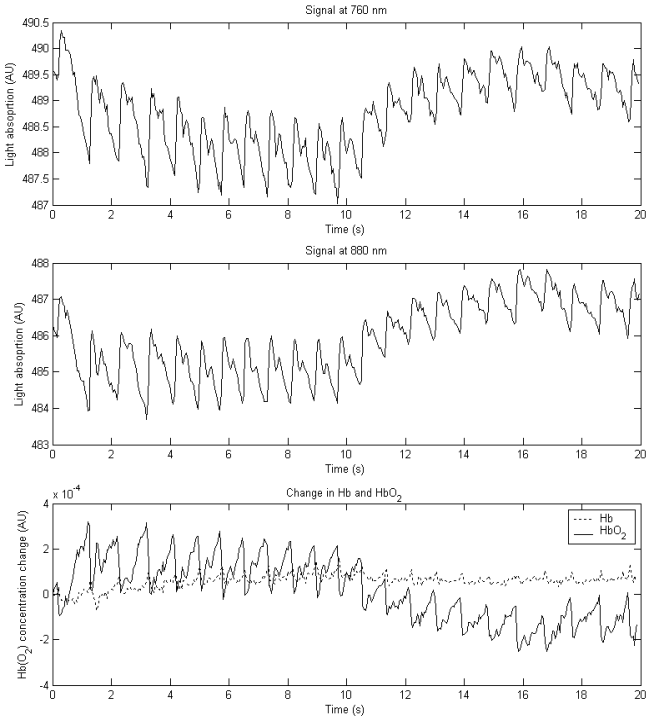

Fig. 3. (Top) PPG signal captured at 760 nm, (middle) PPG captured at 880 nm and (bottom) corresponding changes in concentrations of Hb and HbO_2.

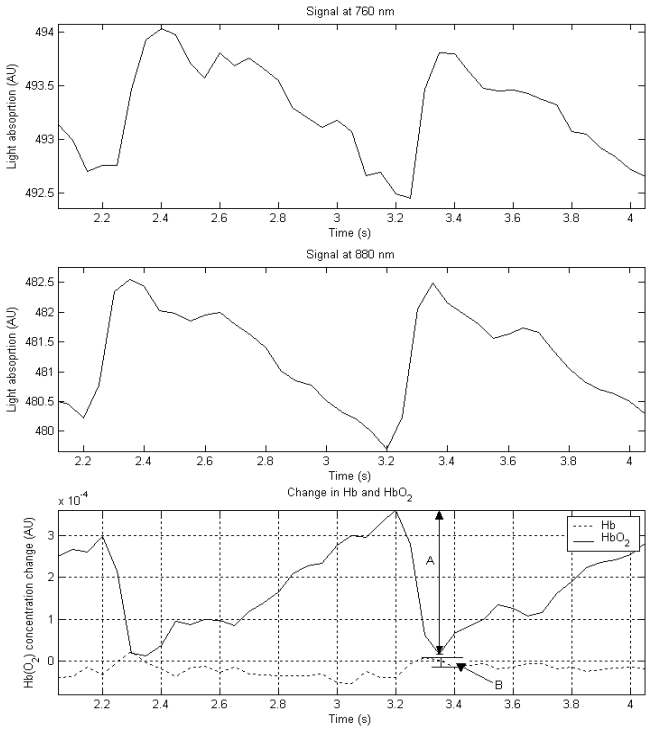

Fig. 4. Calculation of SaO_2 from ΔHbO_2 and ΔHb pulse heights using (5). $3.25\text{x}10^{-4}/(3.25\text{x}10^{-4}+1.75\text{x}10^{-5}) = 94.8\%$.

pulse heights of the Hb and HbO_2 concentration change graphs [10]. Fig. 4 illustrates a calculation of the percentage SaO_2 based on this. The SaO_2 value calculated from the data in Fig. 4 is 94.8%.

IV. DISCUSSION

The SaO_2 value of 94.8% calculated from this data set is lower than would be expected. During the camera-based pulse oximeter testing, a conventional Welch Allyn patient monitoring device was also used to measure the subjects' SaO_2. This device is of the type used in clinical practice. The conventional pulse oximeter recorded an SaO_2 value of 97% throughout the test depicted in Fig. 4. There are several factors that could account for the difference in performance. As stated earlier the detector of a conventional pulse oximeter is relatively shielded from extraneous light. The camera sensor is exposed to light from the diodes as well as ambient light from overhead lighting, monitors and daylight. Commercially available pulse oximeters also differ in two important ways from the system described here. Firstly commercial pulse oximeters employ a look-up table to convert the amplitude ratio of the two wavelengths of light to a saturation percentage. Secondly commercial oximeters use a weighted average of previously calculated SaO_2 values to calculate the current value thus mitigating the effects of spurious values.

Further steps can be taken to reduce the camera-based system's sensitivity to ambient light. The incorporation of a NIR filter before the lens would remove much of the unwanted light from the visible band. The inclusion in the triggering sequence of a dark frame – capturing a frame with

neither diode energized – and subtracting the signals in this frame from the signals in the adjacent frames could also mitigate the effects of ambient light.

V. CONCLUSIONS

A CMOS camera-based system for pulse oximetry imaging in transmission mode has been described and focus brought to bear on its potential application to wound assessment. The system has been shown to be capable of simultaneously capturing PPG signals at two different wavelengths without making contact with the tissue under investigation. The system's vulnerability to ambient light has been highlighted as well as measures that can be taken to improve this. The system in its current transmission mode set up is limited to oximetry imaging of the extremities. In order to be successfully applied to a wound assessment task a reflection mode system must be developed.

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