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IN SITU MONITORING OF MEAN BLOOD OXYGEN SATURATION USING EXTENDED MODIFIED LAMBERT BEER MODEL

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ABSTRACT

We present the use of Extended Modified Lambert–Beer model for optical monitoring of mean blood oxygen saturation (S_mO_2) via a fitting procedure. This work focuses on the absorption characteristics of hemoglobin derivatives in the wavelength range of 520–600 nm to give the best estimates of S_mO_2. The study of the feasibility of applying this analytic method to skin oximetry is via spectroscopy data collected from fingertips of four healthy volunteers both at rest and during arterial blood occlusion condition. The results revealed a decrease in the mean of mean and standard deviation of S_mO_2 value of fingertips from $94.5 \pm 2.19\%$ when volunteers were at rest to $56.76 \pm 5.8\%$ during the arterial blood occlusion measurement. The larger variation in the value estimated for blood occlusion condition could be a result of differences in volunteers' physical fitness and hypertension status. These estimated S_mO_2 values agreed reasonably well with the value reported in most of the previous studies. This work concluded that the proposed technique can potentially be used as a complementary technique to clinical assessment of skin grafts and burnt skin.

Keywords: Extended modified Lambert–Beer model; Mean blood oxygen saturation; Skin oximetry; Spectroscopy.

INTRODUCTION

Oximetry has been used extensively for several decades as one of the most important diagnostic tools to monitor one's blood oxygen saturation or oxyhemoglobin (OxyHb) saturation. This parameter is used to determine the amount of oxygen bound to hemoglobin, and is normally related to the health of a person's cardiovascular system and the recovery rate of an affected area from the injury. The principle of modern pulse oximetry is based on the change in the color of Hb in respond to the oxygen level[1] and this device relies on a look up table to interpret the measured signals into the related arterial

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blood oxygen saturation, SpO_2. It is, however, the ethical issue that prevents manufacturers from introducing severe hypoxia on volunteers during the tabulation of this look up table. Therefore, the main limitation of modern pulse oximetry lies on its poor performance at low SpO_2. Besides the operation of pulse oximetry, required the use of a finger probe to allow transmittance measurement on the selected fingertip, limiting the application of this device to certain skin sites.

The use of continuous wave (CW) spectroscopic technique in reflectance mode for continuous, noncontact and noninvasive measurement of blood oxygen saturation at different skin sites has gained popularity as an alternative to pulse oximetry, especially in clinical assessment of skin grafts and burnt skin. The reflectance data collected from this system are often used to estimate the required oxygen saturation and this strategy does not require a look up table for the interpretation work. The required parameter is obtained by means of fitting using either analytic models or a library of data simulated using Monte Carlo method or diffusion model.[2,3] Amongst the commonly used analytic models are Modified Lambert–Beer law (MLBL) that assumes a linear relationship between light absorption and attenuation. In addition to the fitting method, this parameter is also commonly solved via simultaneous solution of this linear model. The relationship between light attenuation and absorption in blood is, however, a nonlinear and multi-valued function owing to the wavelength dependent scattering properties of different skin layers. This results in poor accuracy in the value estimated by the MLBL. An attenuation model extended from MLBL developed by Huong and Ngu[1] was reported to be able to describe the true light absorption and scattering processes beneath skin surface. The corresponding work, however, lacks experimental work to support its findings.

This paper presents the use of Extended Modified Lambert Beer model to extract the percent blood oxygen saturation using reflectance spectroscopy performed on skin surface. This study assumes that OxyHb and deoxyhemoglobin (Hb) are the only two absorbing components present in blood and a pressure of 180 mmHg applied on upper arm of an individual is able to shut oxygenated blood into the lower arm. The performance of the analytic method proposed in this work is evaluated via the ability of the technique to detect changes in oxyhemoglobin saturation under different experimental settings and by comparison with the results presented in previous works. The focus of this study is on signals measured across wavelengths 520–600 nm as suggested by Huong and Ngu.[1]

MATERIALS AND METHODS

Noncontact Optical System and Data Calibration

The illuminating source used in this system consists of a 9 W white light emitting diode (LED) (Model no. SMD 5730 from Aira Technologies). Light reflected from the sample is collected by an optical fiber placed at an angle of approximately 40° from the source axis in Fig. 1 to prevent detection of specular reflectance signals. The distance between the optical fiber tip and the specimen surface is about 30 mm. The measurement of diffuse reflected light intensity is via a commercial spectrometer (USB4000-VIS-NIR, Ocean Optics Inc., Florida). This spectrometer is connected directly to the optical fiber to produce intensity spectra with resolution of approximately 0.2 nm across wavelength range of 200–850 nm. The detected signals are then transferred and stored to a computer via universal serial bus (USB) port for further offline analysis.

Next, sample intensity data from spectroscopy system are corrected by taking both white and black standard data. While the dark reference data are taken to subtract the effects of system dark noise from the signals and were acquired with a shutter placed in front of the optical fiber, the white standard data are given by the reflectance data from spectralon (from Labsphere, Inc.) with 99% reflectance. Using the captured white and dark reference, the corrected and calibrated light attenuation at a wavelength, λ, $A_{\text{corr}}(\lambda)$, is calculated as:

$$A_{\text{corr}}(\lambda) = \log \frac{I_{\text{white}}(\lambda) - I_{\text{dark}}(\lambda)}{I_{\text{sample}}(\lambda) - I_{\text{dark}}(\lambda)}, \quad (1)$$

where I_{sample}, I_{white} and I_{dark}, respectively, are the light intensity collected from the measurement sample, white and dark standard.

Light reflected from the measurement sample is taken here to be an integration of light encompassing dermal

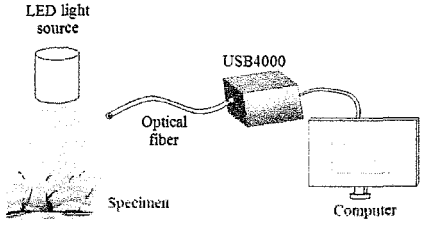

Fig. 1 Noncontact spectroscopy system.

Table 2. Comparison of Mean Blood Oxygen Saturation Estimated for Volunteers at Rest and During Arterial Blood Occlusion Measurement.

No.	Investigator[ref]	Experimental Setting	Reported Mean Blood Oxygen Saturation (%)
1	Caspary *et al.*[5]	Volunteers at rest	92 ± 2.6
2	Zhang *et al.*[6]	Volunteers at rest	93 ± 1
3	Kobayashi *et al.*[7]	Volunteers at rest	68 ± 6
4	Kobayashi *et al.*[7]	Arterial blood occlusion	48
5	Vogel *et al.*[2]	Arterial blood occlusion	35
6	This work[a]	Volunteers at rest	94.5 ± 2.19
7	This work[a]	Arterial blood occlusion	56.76 ± 5.8

[a]Mean of $\langle S_mO_2 \rangle$ of the recruited volunteers in Table 1 for at rest and arterial blood occlusion experiment.

DISCUSSION

The S_mO_2 values shown in Fig. 3 are the mean blood oxygen saturation estimated for arteries, veins and capillaries in skin of volunteers. The mean of $\langle S_mO_2 \rangle$ value of 94.5% given from the value predicted for different fingertips of all volunteers when they were at rest condition in Fig. 3(A) is similar to that reported by Caspary *et al.*[5] and Zhang *et al.*[6] in Table 2. The lower blood oxygen saturation value reported by Kobayashi *et al.*[7] estimated for volunteers at rest could be a result of inappropriate assumptions made on skin thickness and scattering related parameters in their Monte Carlo model. The high S_mO_2 values estimated for fingertips in this work is caused by the presence of arteriovenous anastomoses (AVAs), which are available in abundance at the acral (e.g. palm of hand, feet and face) than at the nonacral skin (e.g. back of the hand). The AVAs connect arterioles to venules and produce shunting of the arterial blood into the venous compartment via AVAs during the muscle relaxation measurements. In addition, Thorn *et al.*[8] suggested that high S_mO_2 observed at the acral area is caused by the high cutaneous blood flow that increases the washout of Hb. This high S_mO_2 value of fingertips is relatively consistent for all volunteers and digits with low mean standard deviation of S_mO_2 of 2.19% in Table 2 indicating a good repeatability of the measurement.

For the arterial blood occlusion case, arteries are closed and the supply of the oxygenated blood into the lower arm is shut, therefore the arterial blood in the venous compartment is low in oxygen. This reduces the S_mO_2 value in all the collected samples with mean $\langle S_mO_2 \rangle$ value of 56.76% in Table 2. These trends are similar to that observed by Vogel *et al.*[2] and Kobayashi *et al.*[7] in Table 2. Larger variation in the S_mO_2 value with mean standard deviation of S_mO_2 of 5.8% observed in the arterial blood occlusion case is likely due to differences in volunteers' physical fitness and hypertension status. This work observed the largest standard deviation of 7.44% for S_mO_2 predicted for different fingertips of volunteer B during arterial blood occlusion experiment shown in Table 1. Since measurement on each fingertip is carried out at different point of time, the fluctuation in the estimated S_mO_2 value for different fingertips of each individual is most probably linked to vasomotion induced temporal variation of blood oxygen saturation, which is also observed in the work by Caspary *et al.*[5] and Thorn *et al.*[8] There is ongoing work on real-time monitoring of S_mO_2 to confirm this.

This work assumes OxyHb and Hb are the only absorbers, thus the light absorption by melanin was deliberately minimized by working on the wavelength range where its absorptivity is low (i.e. wavelength range of 520–600 nm). Even though this will not eliminate the absorption of light by the melanin entirely, the changes in the S_mO_2 value with different sets of performed measurements match reasonably well with most of the results reported by other investigators. A possible explanation of this is that the absorption of melanin decreases monotonically with the wavelength, and this trend of curve is in good agreement with the $\mu_a^*(\lambda)$ of human skin as discussed in Huong and Ngu.[4] The curve fitting model in Eq. (7) includes a linear λ term to take into account this monotonic decrease in attenuation due to the absorption by melanin and scattering processes.

CONCLUSION

In summary, this work has demonstrated the performance of Extended Modified Lambert–Beer model at detecting changes in S_mO_2 using measurements on fingertips when different actions are taken. The S_mO_2 values presented in this work can only serve as a reference as these values may vary among the skin colors, human populations, physical condition of a person and the location of the blood vessels. The value may also be varied with one's vasomotion mechanism and vascular microcirculation. Further work is currently underway to investigate these hypotheses. The proposed stategy can also be modified to eliminate the effects of skin thickness and epidermal absorption on the measurement data to allow a high consistency in the estimated S_mO_2 value across different skin regions. Even though the results shown in this study have preliminary characteristics, they reveal several promising applications where the technique described in this work can be used. The proposed strategy allows noncontact monitoring of one's

mean blood oxygen saturation, hence can potentially be used for clinical assessment of skin grafts and burnt skin.

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