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Multispectral Imaging System for Quantitative Assessment of Transcutaneous Blood Oxygen Saturation

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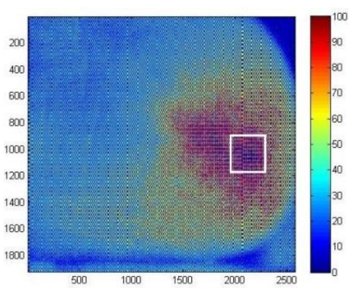
Graphical abstract



Abstract

This paper presents the use of Extended Modified Lambert Beer (EMLB) model for quantification of transcutaneous blood oxygen saturation (S_tO_2) via a noninvasive approach. Continuous wave (CW) reflectance spectroscopy system is employed for measurement of intensity reflected from left index finger of an Asian nonsmoking volunteer at resting condition. Multispectral images captured in the wavelength range of 520 − 600 nm at an interval of 10 nm are mathematically analyzed and fitted using the developed fitting algorithm to give the best estimation of S_tO_2. The result from this preliminary study revealed a mean S_tO_2 value of 75 ± 5% for the participating individual, which value agreed considerably well with that presented in previous works. This work concluded that the developed spectroscopy system and quantification technique can potentially be used as an alternative means to clinical assessment of wound healing progress.

Keywords: Transcutaneous blood oxygen saturation; Extended Modified Lambert Beer model; reflectance spectroscopy; skin oximetry

Abstrak

Kertas kerja ini membentangkan penggunaan *Extended Modified Lambert Beer (EMLB) model* untuk kuantifikasi ketepuan oksigen darah *transcutaneous* (S_tO_2) melalui pendekatan bukan invasif. Sistem spektroskopi jenis gelombang berterusan digunakan untuk mengukur intensiti dari jari telunjuk kiri seorang perokok Asian dalam keadaan rehat. Imej *multispectral* yang diambil dalam julat panjang gelombang antara 520 − 600 nm pada selang 10 nm dianalisis menggunakan fitting algorithm untuk memberikan anggaran terbaik S_tO_2. Hasil kajian awal menunjukkan nilai purata S_tO_2 75 ± 5% untuk individu tersebut, selaras dengan nilai yang dibentangkan dalam kajian sebelumnya. Secara kesimpulan, sistem spektroskopi dan teknik kuantifikasi yang dibangunkan berpotensi sebagai alternatif kepada penilaian klinikal penyembuhan luka.

Kata kunci: Ketepuan oksigen darah *transcutaneous*; *Extended Modified Lambert Beer model*; pantulan spektroskopi; oximetri kulit

1.0 INTRODUCTION

For years, the use and application of reflectance spectroscopy are much discussed subjects in biomedical optic researches. This is owing to the simplicity of the system and its noninvasive nature. Many of the researches focus on using the corresponding system to characterize and determine optical properties of biological tissues [1]. Noninvasive and noncontact measurement of blood oxygen saturation (SO_2) for the application of wound healing assessment has gained an increasing interest in the medical arena [1,2]. This is in accordance to the major drawbacks of pulse oximeter, which clinical measurement means is via the use of a finger clip that could shred skin tissue. In addition, the accuracy of a pulse oximeter is limited to SO_2 value of greater than 70% [3]. Meanwhile, readings taken from a patient suffering from cyanosis, a condition in which blood lacks oxygenated hemoglobin, may result as incongruous given that pulse oximeter detect mostly plasma in blocked microcirculatory tissue [4].

Spectroscopy imaging is a technique commonly practiced nowadays in medical diagnosis and prognosis to provide intensity information of a sample across a selected wavelength range. Spectral measurement from both skin surface and below skin surface is obtained when light is sufficiently reflected, transmitted, and absorbed by the skin and tissues. An advantage to this technique is its noninvasive attribute that complemented the technique as an ideal wound oximeter. Different techniques have been developed to provide detailed information regarding the composition of oxygen level beneath the human skin based on the obtained spectroscopic data. Amongst these techniques are the use of either a library of data simulated using either Monte Carlo method or diffusion model [5], or analytical model such as Modified Lambert Beer (MLB) model [6, 7], Extended Modified Lambert Beer (EMLB) model [3], cubic model or a nonlinear fitting model [8].

Continuous wave (CW) reflectance spectroscopy system used for noninvasive estimation of one's transcutaneous blood oxygen saturation, S_tO_2, are commonly categorized into optical point spectroscopy [9] and imaging spectroscopy such as multispectral [5] and hyperspectral imaging [10]. These approaches are able to produce the spectral information of skin tissues. In contrary to point spectroscopy technique, multispectral imaging technique is able to detect intensity of the light at wider image coverage. Even though hyperspectral imaging system is able to provide hundreds to thousands of spectral bands, it has a longer data acquisition time compared to that of multispectral system that captured images at a relatively discrete and narrow band [11].

In this work, we employed Extended Modified Lambert Beer model developed by Huong and Ng [3] for quantification of one's S_tO_2 level using experimental data collected from the developed multispectral imaging system. This paper is arranged as follows: Section 2 described the developed system, experiment technique and quantification strategy. This work used extinction coefficient of light absorbing components in the wavelength range of 520 − 600 nm for analysis, and we assume hemoglobin, Hb and oxyhemoglobin, HbO_2, are the only absorbing medium in blood. The S_tO_2 result obtained from this work is presented and compared with that reported in the literature in section 3 followed by a discussion of the result and conclusion in section 4.

2.0 MATERIALS AND METHOD

2.1 Multispectral Imaging System

Figure 1 shows the optical arrangement of the multispectral imaging system used in this work. This work employed reflectance spectroscopy system for measurement of intensity reflected from skin sites. The illuminating source used in this system consists of a high intensity light emitting diode (LED) (XLamp® XQ-E LED from Cree) that shone onto a monochromator (Oriel Mini Monochromator model 78026 from Newport). Light focused at the entrance slit (slit size 30μm × 4mm) of the monochromatic system is diffracted by a diffraction mirror placed within the monochromator, and images within the wavelength range of 520 − 600 nm are collected at an interval of 10 nm. The reason for the selection of this wavelength range is due to the distinctive absorption of hemoglobin derivatives in the dermis layer [12]. This system used a cooled charge-coupled device (CCD) camera (BUC4-500C Cooled CCD Digital Camera from BestScope) to detect the reflected light. A plano-convex lens (with diameter, Ø = 12.7 mm and focal length, f_L= 50.2 mm) is placed in front of the CCD camera, to focus reflected light onto the CCD.

Prior to the experiment, the volunteer gave written, informed consent and was briefed on the experimental procedure before the measurements. These measurements were conducted in a dark room. Measurements were taken after the volunteer has been acclimatized to room temperature for ten minutes. During measurements, the volunteer was asked to sit in an upright position and to breathe normally with the elbow placed lower than the heart level to allow unimpeded blood flow. Measurement of white reference data is taken from the reflectance of a spectralon (from Labsphere, Inc.) with 99% reflectance whilst the dark reference data is obtained by covering the CCD with a shutter cap.

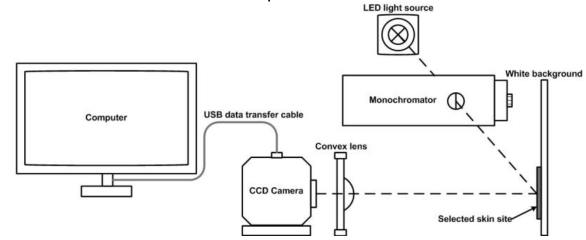


Figure 1 Optical arrangement of the employed multispectral imaging system

The calibrated light attenuation at corresponding wavelengths, $A_{corr}(\lambda)$ is given by:

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

where I_{sample}, I_{white} and I_{dark} are the light intensity of the image sample, white and dark reference, respectively, retrieved from the multispectral imaging system.

2.2 Extended Modified Lambert Beer Model

In this work, percent S_tO_2 is estimated using EMLB model proposed by Huong and Ngu [3] shown in Equation 2.

$$A(\lambda) = G_0 + \mu_a d_0 + G_1\lambda + \lambda \exp(-\mu_a d_1) \quad (2)$$

This fitting model is previously extended from the MLB model that is first introduced by Duling and Pittman [13]. The light absorption is represented by μ_a while parameter d_0 in Eq. 2 is normally taken as "light pathlength". Parameters G_0 and G_1 in Eq. 2 represent light attenuation offset and light attenuation due to scattering and epidermal light absorption, respectively. The nonlinear term in Eq. 2 is used to represent intertwined effects of light absorption and scattering processes in dermal layer [13, 14].

The total light absorption, μ_a, is defined as the sum of the product of concentration, C and wavelength dependent extinction coefficient, $\varepsilon(\lambda)$.

$$\mu_a = \varepsilon_{OxyHb}(\lambda) C_{OxyHb} + \varepsilon_{Hb}(\lambda) C_{Hb} \quad (3)$$

Considering blood OxyHb and Hb as the only light absorbers in the dermis layer, the total hemoglobin concentration is given as $T = C_{OxyHb} + C_{Hb}$. Substituting parameter T into Eq 4, light absorption, μ_a is defined as:

$$\mu_a = \varepsilon_{OxyHb}(\lambda) C_{OxyHb} + \varepsilon_{Hb}(\lambda)(T - C_{OxyHb}) \quad (4)$$

The percent S_tO_2 can be expressed as a function of T as [3, 10, 14]:

$$S_tO_2 = \frac{C_{OxyHb}}{T} \quad (5)$$

Using Eq. 5, Eq. 4 can be further rearranged to give:

$$\mu_a = \left(\left(\varepsilon_{OxyHb}(\lambda) - \varepsilon_{Hb}(\lambda) \right) S_tO_2 + \varepsilon_{Hb}(\lambda) \right) T \quad (6)$$

Given that $\Delta\varepsilon$ denotes $\varepsilon_{OxyHb}(\lambda) - \varepsilon_{Hb}(\lambda)$, the EMLB model in Eq. 2 can also be expressed as followed:

$$A(\lambda) = G_0 + \left(\Delta\varepsilon S_tO_2 + \varepsilon_{Hb}(\lambda) \right) d_0 T + G_1\lambda + \lambda \exp\left(-\left(\Delta\varepsilon S_tO_2 + \varepsilon_{Hb}(\lambda) \right) d_1 T \right) \quad (7)$$

The S_tO_2 in Eq. 7 is also known as mean blood oxygen saturation, and is the blood oxygen saturation value encompassed across the arteries, veins, capillaries and skin tissues.

2.3 Iterative Fitting Procedure

The attenuation data calculated using the intensity data given from the multispectral system in Figure 1 is fitted using the EMLB model in Eq. 7. We used *fminsearch* function available in MATLAB to solve the optimal value of nonlinear unknown parameters (i.e. S_tO_2, G_0, d_0T, G_1, d_1T) in Eq. 7. This fitting algorithm applied unconstrained nonlinear optimization routine to iteratively seek the new value of these fitting parameters based on the size of error between the attenuation value from the EMLB model and the measured attenuation value, ΔE. This process began with assignment of initial value of '1' to these unknown parameters. The fitting process is terminated when either the mean of ΔE value is less than 1×10^{-12} or the number of iteration has achieved 1000, when the optimal S_tO_2 value is assumed to have been obtained.

3.0 RESULTS AND DISCUSSION

Figure 2 shows the S_tO_2 map estimated based on the reflectance data collected from the left index finger of the recruited volunteer at resting condition. The overall mean of S_tO_2 shown in Figure 2 is calculated as 53.57%. Meanwhile the highest mean S_tO_2 of 75% is found in the region indicated by box in Figure 2. Since there is currently no gold standard with which the measured S_tO_2 can be compared to, the obtained result is compared with that reported in previous works in Table 1.

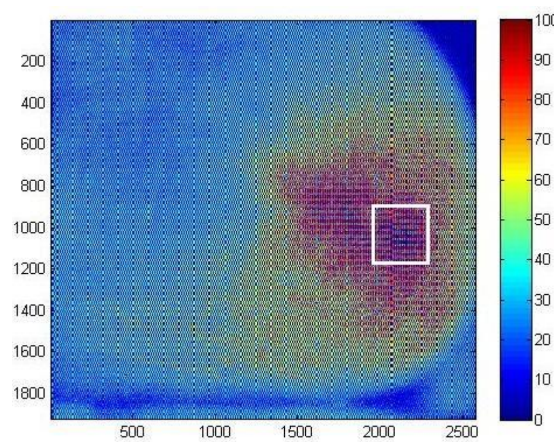


Figure 2 The transcutaneous blood oxygen saturation map, S_tO_2, calculated for the index finger of the recruited subject at resting condition

Table 1 Comparison of transcutaneous blood oxygen saturation, S_tO_2, estimated for nonsmoker in this work and in literature

No.	Investigator	Analytical model	Reported mean blood oxygen saturation
1	Kobayashi *et. al*[a]	Cubic function	68 ± 6%
2	Thorn *et. al*[b]	Fitting via MLB model	63 ± 11%
3	Vogel *et. al*[c]	Power law model	60%
4	A. Huong[d]	Improved linear equation and nonlinear fitting model, Gradient processing technique	74 ± 7%
5	This work	Fitting via EMLB model	75 ± 5%

[a]M. Kobayashi, Y. Ito, N. Sakauchi, I. Oda, I. Konishi, and Y. Tsunazawa, "Analysis of nonlinear relation for skin hemoglobin imaging," Optics Express, vol. 9, pp. 802-812, 2001.
[b]C. E. Thorn, S. J. Matcher, I. V. Meglinski, and A. C. Shore, "Is mean blood saturation a useful marker of tissue oxygenation?," American Journal of Physiology-Heart and Circulatory Physiology, vol. 296, pp. H1289-H1295, 2009.
[c]A. Vogel, V. V. Chernomordik, S. G. Demos, R. Pursley, R. F. Little, Y. Tao, et al., "Using noninvasive multispectral imaging to quantitatively assess tissue vasculature," Journal of Biomedical Optics, vol. 12, pp. 051604-051604-13, 2007.
[d]A. K. C. Huong, "Spectroscopic analysis of scattering media via different quantification techniques," Thesis, University of Nottingham, 2012.

Referring to the result shown in Figure 2, this study observed the highest S_tO_2 value of 75 ± 5% (indicated by box) at the fingertip of the recruited participant. This is likely owing to the abundance of arteriovenous anastomoses (AVA) in this skin region that connects arterioles to venules when the muscle is in relax condition [15]. This high S_tO_2 calculated with the volunteer is at resting condition is close to that reported by A. Huong [10] when experiments were performed on palm area. The latter employed three different models listed in Table 1 and has demonstrated consistency in the estimated S_tO_2 value. Meanwhile, the slightly lower S_tO_2 reading observed in the work by Thorn *et al.* [15] in Table 1 is likely due to the limitations of MLB model as discussed in the work of A. Huong and X. Ngu [3]. A similar range of S_tO_2 value is also observed in the works of Kobayashi *et al.* [6], who used cubic function along with a multi-wavelength system for analysis of skin spectra. Inappropriate assumption of skin thickness and light propagation across different skin layers used in the Monte Carlo simulation presented in the work by Vogel *et al.* [5] in Table 1 could also result in differences in these calculated value.

This work shows that the multispectral images collected for wavelength range of 520 − 600 nm at step resolution of 10nm obtained from the developed system is able to be quantified and used to deduce one's S_tO_2. This is likely due to the significant differences in the absorptivity of hemoglobin components in this wavelength range as discussed in the work of A Huong [3]. We have shown the feasibility of using the developed system and the analytic technique for noncontact monitoring of local changes in S_tO_2 during wound healing progression. This is in accordance to the work of L. Zhu [2] who observed significant changes in HbO_2 concentration with proliferation of newly formed capillaries in the dermis layer during the early stages of healing.

4.0 CONCLUSION

This paper has demonstrated the application of a noninvasive multispectral imaging system analyzed via Extended Modified Lambert Beer model for noninvasive estimation of one's percent S_tO_2. The developed system and strategy observed the highest percent S_tO_2 of 75 ± 5% at index finger of a nonsmoking individual. This work concluded the corresponding system and technique could potentially be used for optical monitoring of wound healing rate in response to different medical treatment.

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