A Low-cost CCD-based Imager for Mapping Venous Oxygenation

Jun Li and Xiao Zhang
Centre for Optical and Electromagnetic Research
South China Normal University, Guangzhou 510006, China

Abstract— A low-cost imaging system is developed for mapping skin venous oxygen saturation. The system consists of a black-and-white CCD camera and cheap LED light sources working at 660 nm and 800 nm wavelength. With the two wavelength sources alternately illuminating the skin, the CCD camera collects images carrying signals of diffuse reflectance before and after the venous occlusion induced by a low cuff pressure. Using the Monte Carlo method, the relationship between the venous saturation value and the diffuse reflectance is determined. Therefore images captured can be converted into the venous saturation map. Experiments on human fingers and palm show that the measured data is in line with previous published data, demonstrating the effectiveness of the system on monitoring blood oxygenation in local tissues.

1. INTRODUCTION

Diffuse reflectance spectroscopy with multi-wavelength can be used for measuring tissue oxygenation; however it generally works in a tissue-optode contact manner and provides point measurement. The need for non-contact mapping of skin tissue oxygenation in some clinical environments facilitates the development of a CCD-based imaging system [1–4].

Pulse oximetry measures the arterial oxygen saturation by monitoring the pulsatile alterations in light transmission of two or more wavelengths due to change in arterial blood volume induced by cardiac cycles. It provides arterial blood oxygen saturation S_aO_2, a global parameter mainly reflecting cardiac information and lung function. Similarly, by varying the venous blood volume and recording changes in light transmission (or reflection), venous oxygen saturation S_vO_2 can also be obtained [5]. In contrast to arterial oxygen saturation, venous oxygen saturation provides information on local oxygen level after perfusion and tissue uptake, which is closely associated with the local tissue health. Thus venous oximetry can be applied in clinics to quantitatively monitor wounds, ulcers and oxygen metabolism in ischemic tissues.

The author reported earlier a CCD-based imaging system for non-contact mapping of venous oxygen saturation [3, 4]. The model used for converting the diffuse reflectance to S_vO_2 is a hybrid model combining the Beer-Lambert law and the photon diffusion theory. However, the diffusion approximation is not valid in the case where the optical path length is very short, for example, emitted light of a pixel coming from very adjacent region around the pixel. This may result in some unpredictable errors for the calculated S_vO_2. Therefore, in this work, we use Monte Carlo simulation [6–8] to investigate the relationship between S_vO_2 and the diffuse reflectance. In addition to the theoretical model, hardware improvements are also made for the imaging system to achieve high quality of images. To demonstrate the effectiveness of the imaging system, time series of images were taken from fingers and palm of a human subject. The images recorded were converted into S_vO_2, and two-dimensional (2-D) S_vO_2 maps are also presented.

2. METHODS

2.1. Imaging Setup

The system (as shown in Figure 1) is made up of a 14-bit black-and-white CCD camera (Pike F-032, Allied Vision Technologies GmbH, Stadtroda, Germany) and two high-power LED sources with one working at 660 nm and the other at 800 nm wavelength. The illuminating light is transmitted through a fiber optic ring light guide to insure a uniform illumination on the skin. To suppress direct surface reflection which carries useless information on tissue oxygenation, cross polarization detection is used. During image acquisition, the two LED sources illuminate alternately, which is controlled by a custom electronics and software developed with Labview. The CCD camera is set to work in a trigger mode, thus the onset of the illuminating light and the shutter of the camera is synchronized.

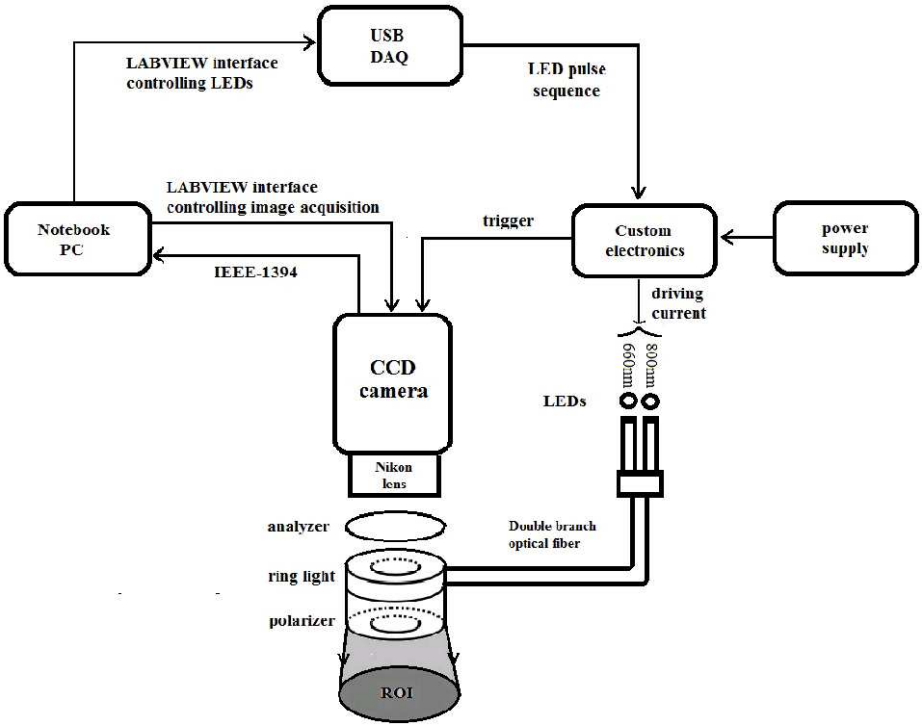

Figure 1: Block diagram for the imaging system.

2.2. Theoretical Model and Monte Carlo Simulation

Based on either the Beer-Lambert law or the hybrid model, there exists a linear relationship between S_vO_2 and the ratio-to-ratio R, $R = (\Delta I/I)_{\lambda_1}/(\Delta I/I)_{\lambda_2}$ is the ratio of the relative intensity change for the two wavelengths, a measurable quantity. However, the relationship derived previously, e.g., from the hybrid model, may not accurate enough for the back reflection geometry with the planar wave illumination, thus Monte Carlo simulation is used to investigate the relationship between S_vO_2 and R.

Monte Carlo simulation describes accurately light propagation in a turbid medium without any prior approximation [6–8]. To use the ratio-to-ratio method to calculate S_vO_2, two questions at least have to be addressed. First, with a certain S_vO_2, is the R a constant over a range of venous volume alteration? If the answer is yes, which implies there is a one-to-one relationship between R and S_vO_2, then the second question need to be answered, what is the exact relationship between R and S_vO_2. We use Monte Carlo simulation to investigate these issues.

Unlike the photon diffusion model, the Monte Carlo simulation can treat the tissue as a multi-layered medium without any restriction on either optical or geometric parameters. In our simulation, we used two skin models, one is a semi-infinite homogeneous model, and the other is a two-layered model. Assume the refractive index for skin tissue is approximately same for the two wavelengths, $n = 1.43$; thickness of epidermis, $d\_\text{epi} = 50\,\mu\text{m}$; volume fraction of melanin in epidermis, $f\_\text{mel} = 2.0\%$; volume fraction of blood in dermis and tissue, $f\_\text{blood} = 3\%$; average tissue blood oxygen saturation $S_tO_2 = 74\%$. The extinction coefficients of the oxygenated hemoglobin (HbO) and deoxygenated hemoglobin (Hb) for the two wavelengths are taken from the published data [9]. With the above data, the optical parameters used for the models can be estimated [10] and listed in Table 1.

Table 1: Optical parameters used for the Monte Carlo simulation.

Optical Parameters	μ_a [660 nm] (/cm)	μ_a [800 nm] (/cm)	μ_s' [660 nm] (/cm)	μ_s' [800 nm] (/cm)
		Semi-infinite model		
	0.718	0.477	22.8	16.4
		Two-layer model		
Epidermis	5.668	3.084	22.8	16.4
Dermis	0.459	0.377	22.8	16.4

Figure 2 gives the simulation result for R with respect to the alteration in venous volume. Clearly one can see for both models, the R is nearly constant over a wide range of venous volume alteration, e.g., from 6% to 50% volume change. This means R does not depend on the venous volume change as long as the venous volume change is within 6%–50%, implying the ratio-to-ratio method may work in this range for estimating S_vO_2.

The second question was investigated by varying the venous blood volume (e.g., a 30% increase in volume) for different levels of S_vO_2 to induce changes in back reflectance for the two wavelengths. Therefore for each S_vO_2 a R value can be obtained. The simulation result is shown in Figure 3. By the polynomial fitting, the analytical relationship between the R and S_vO_2 can be identified for the two skin models. For the semi-infinite model, $S_vO_2 = -0.0304R^2 - 0.2031R + 0.9474$, while for the two-layered mode, $S_vO_2 = -0.0652R^2 - 0.1593R + 1.0084$. In contrast to the linear relationship between R and S_vO_2 derived from either Beer-Lambert law or the hybrid model, the relationship

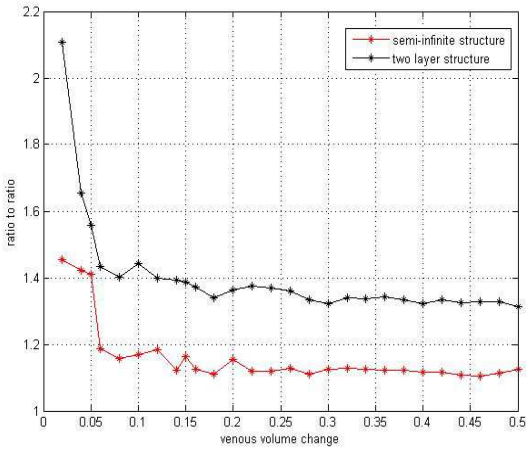

Figure 2: Ratio-to-ratio R with respect to changes on venous blood volume for the two skin models.

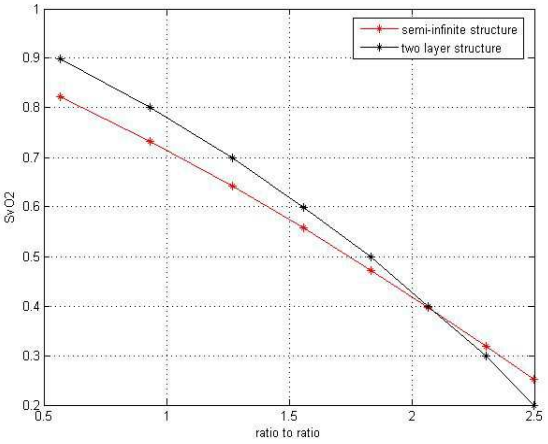

Figure 3: S_vO_2 with respect to R for the two skin models.

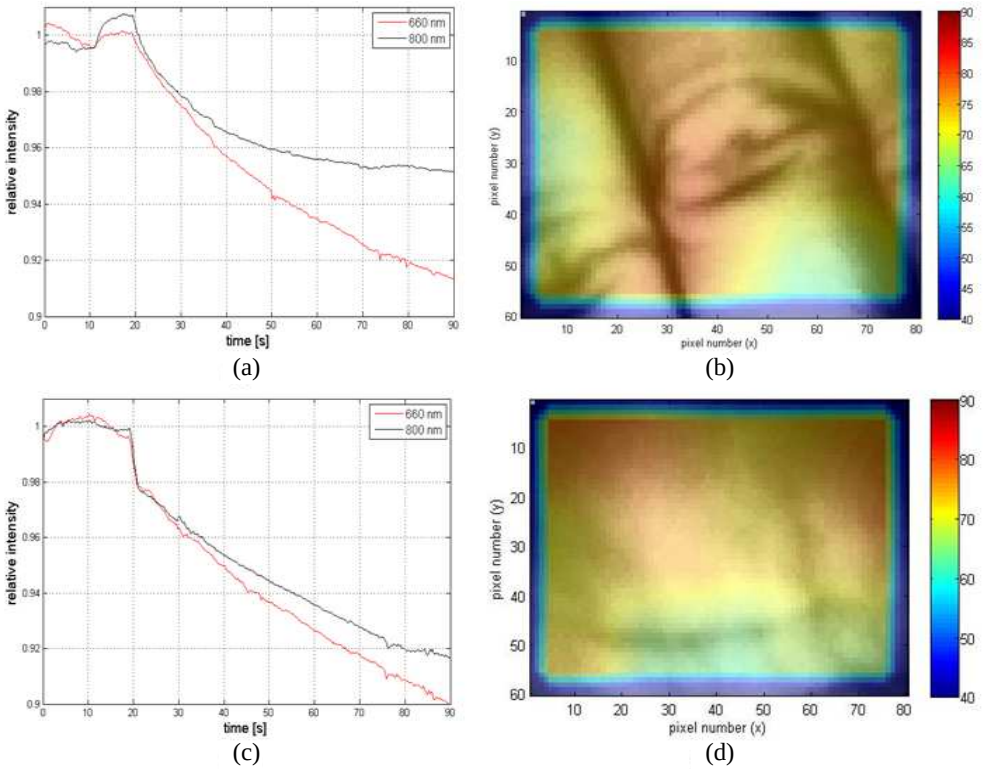

Figure 4: Relative intensity change over time for fingers (a) and palm (c), and SvO_2 map for fingers (b) and palm (d). Venous occlusion occurs at time $t = 20$ s.

identified from the Monte Carlo simulation is no longer linear, especially for the two-layered model with higher non-linearity.

In comparison between the two skin models, the semi-infinite model gives smaller value as R is less than 2.1, slightly larger value as R is bigger than 2.1. However, within a wide reasonable range of S_vO_2, e.g., 30%–85%, the maximum difference in the estimated S_vO_2 between these two models is only 5–6% in amplitude.

3. EXPERIMENTAL RESULT AND DISCUSSION

A preliminary experiment was performed with the developed imaging system. To induce venous volume change, a low cuff pressure (∼50 mmHg) was applied on a subject wrist. This pressure is high enough to occlude the vein with less effect on the arterial flow. Dynamic images were collected from fingers and palm of the subject before and after the occlusion, separately.

Figure 4 shows the result of measurements on the fingers [Figures 4(a)–(b)] and the palm [Figures 4(c)–(d)]. The curves in Figure 4(a) give the average change in intensity for the red and IR light over the entire image area. The pressure cuff was quickly inflated to 50 mmHg at time $t = 20$ s to induce a rapid venous occlusion. Both 660 nm and 800 nm curves decrease rapidly due to the increase in light absorption caused by venous blood accumulation after the vein was occluded.

Using the two-layered model, images of diffuse reflectance are converted into the venous saturation map, as shown in Figure 4(b). It is a 2-D color map of S_vO_2 overlaid on the grayscale image of the fingers. The average S_vO_2 in the mapping region is $70.6 \pm 7.2\%$. The time window used for calculating palm S_vO_2 is 20–50 s.

Results for the palm are shown in Figures 4(c) and (d), the reflectance of 800 nm does not reach the saturation within the measurement period, possibly because it could be longer before the venous blood stops accumulating in the palm than in the fingers. The average S_vO_2 in the mapping region [Figure 4(d)] is $71.4 \pm 7.3\%$. The average values of S_vO_2 for the fingers and palm are in line with the published data [5].

4. CONCLUSION

We develop a low-cost imaging system based on a 14-bit black-and-white CCD camera and high power LED sources for mapping venous oxygen saturation S_vO_2 in skin tissue. Preliminary experiment results show the measured S_vO_2 is consistent with the published data, which demonstrates the effectiveness of the system.

ACKNOWLEDGMENT

This work was supported by Guangdong (China) Innovative Research Team Program (No. 201001D-0104799318).

REFERENCES

1. Wieringa, F. P., F. Mastik, and A. F. W. Van der Steen, "Contactless multiple wavelength photoplethysmographic imaging: A first step toward 'S_pO_2 camera' technology," *Ann. Biomed. Eng.*, Vol. 33, 1034–1041, 2005.
2. Moza, R., J. M. Dimaio, and J. Melendez, "Deep-tissue dynamic monitoring of decubitus ulcers: Wound care and assessment," *IEEE Eng. Med. Biol. Mag.*, Vol. 29, No. 2, 71–77, 2010.
3. Li, J., B. Dunmire, W. Beach, and F. Leotta, "A reflectance model for non-contact mapping of venous oxygen saturation using a CCD camera," *Opt. Commun.*, Vol. 308, 78–84, 2013.
4. Li, J., B. Dunmire, and D. Leotta, "Venous oxygenation mapping with a CCD camera," *2012 Asia Communications and Photonics Conference (ACP)*, 1–3, 2012.
5. Nitzan, M., A. Babchenko, B. Khanokh, and H. Taitelbaum, "Measurement of oxygen saturation in venous blood by dynamic near infrared spectroscopy," *J. Biomed. Opt.*, Vol. 5, 155–162, 2000.
6. Prahl, S. A., M. Keijzer, S. L. Jacques, and A. J. Welch, "A Monte Carlo model of light propagation in tissue," *SPIE Proceedings of Dosimetry of Laser Radiation in Medicine and Biology*, Vol. 5, 102–111, 1989.
7. Wang, L. H., S. L. Jacques, and L. Zheng, "MCML-Monte Carlo modeling of light transport in multi-layered tissues," *Comput. Meth. Prog. Bio.*, Vol. 47, 131–146, 1995.
8. Wang, L. H., S. L. Jacques, and L. Zheng, "CONV — Convolution for responses to a finite diameter photon beam incident on multilayered tissues," *Comput. Meth. Prog. Bio.*, Vol. 54, 141–150, 1997.

9. Zijlstra, W. G. and A. Buursma, "Spectrophotometry of hemoglobin: Absorption spectra of bovine oxyhemoglobin, deoxyhemoglobin, carboxyhemoglobin, and methemoglobin," *Comp. Biochem. Physiol.*, Vol. 118B, 743–749, 1997.
10. Yudovsky, D., A. Nouvong, K. Schomacker, et al., "Two-layer optical model of skin for early, non-invasive detection of wound development on the diabetic foot," *Proc. of SPIE*, 755514-10, 2010.