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An upgraded camera-based imaging system for mapping venous blood oxygenation in human skin tissue

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ABSTRACT

A camera-based imaging system was previously developed for mapping venous blood oxygenation in human skin. However, several limitations were realized in later applications, which could lead to either significant bias in the estimated oxygen saturation value or poor spatial resolution in the map of the oxygen saturation. To overcome these issues, an upgraded system was developed using improved modeling and image processing algorithms. In the modeling, Monte Carlo (MC) simulation was used to verify the effectiveness of the ratio-to-ratio method for semi-infinite and two-layer skin models, and then the relationship between the venous oxygen saturation and the ratio-to-ratio was determined. The improved image processing algorithms included surface curvature correction and motion compensation. The curvature correction is necessary when the imaged skin surface is uneven. The motion compensation is critical for the imaging system because surface motion is inevitable when the venous volume alteration is induced by cuff inflation. In addition to the modeling and image processing algorithms in the upgraded system, a ring light guide was used to achieve perpendicular and uniform incidence of light. Cross-polarization detection was also adopted to suppress surface specular reflection. The upgraded system was applied to mapping of venous oxygen saturation in the palm, opisthenar and forearm of human subjects. The spatial resolution of the oxygenation map achieved is much better than that of the original system. In addition, the mean values of the venous oxygen saturation for the three locations were verified with a commercial near-infrared spectroscopy system and were consistent with previously published data.

1. Introduction

There is an increasing clinical need for the assessment of skin tissue blood oxygenation, with applications such as monitoring ischemia tissue caused by various vascular diseases [1,2], evaluating wound progressing or healing process [3,4] and measuring tissue physiological changes in response to various treatments [5]. To meet this clinical requirement, various optical techniques were developed [6–20], among which camera-based modalities [12–20] are of inherent advantage over those providing only point measurements. Camera-based techniques can not only provide non-contact measurement of the tissue oxygenation, and thus have no risk of infection, but they also cover a relatively large area of skin for clinical assessment.

Multispectral imaging (MSI), originally developed for remote sensing from satellite or aircraft, has been applied to the assessment of skin tissue oxygenation [12–15,17,18]. This technique collects and analyzes a series of images illuminated by different wavelengths of light to obtain the concentrations of various chromophores in the imaged object. When applying MSI to human skin in-vivo, it is able to measure oxygenated hemoglobin (HbO) and deoxygenated hemoglobin (Hb) concentration in the tissue, thereby providing measurement of tissue oxygen saturation (S_tO_2). In fact, S_tO_2 is a weighted average of the arterial blood oxygenation (S_aO_2) and venous blood oxygenation (S_vO_2). As an indicator of oxygen level in the arterial blood, S_aO_2 is a global variable, reflecting lung function rather than local tissue oxygenation. S_vO_2, on the other hand, represents local tissue oxygenation by displaying the remaining equilibrium oxygen saturation in the blood after the tissue is supplied. Therefore S_vO_2 can serve as a direct indicator of local tissue health.

In recent years, techniques for directly measuring S_vO_2 have been developed [6–9,19]. The approach used is very similar to pulse oximetry, except that the measurements are based on venous blood volume alteration. The algorithm used in pulse

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oximetry is based on the measurement of the ratio-to-ratio R, i.e., the ratio of the relative light intensity change for the two working wavelengths. R depends only on the blood oxygenation level, which means there is a one-to-one relationship between R and the blood oxygen saturation. The venous blood volume alteration can be induced naturally with respiration [9], or through artificial means such as applying a blood pressure cuff to occlude the blood return [6–9,19]. However, all methods presented in literature for measuring S_vO_2 were limited to point measurements, until very recently a camera-based imaging system was developed, which could provide the venous oxygenation map [19]. In the system, two wavelengths (660 nm and 800 nm) were used to illuminate the skin in an alternating manner. The venous blood volume change was induced by applying a low cuff pressure. A hybrid model based on the Beer–Lambert law and light diffusion theory was developed to model the diffuse back reflectance with a planar wave illumination. We tested this system on various locations including finger, palm and arm for different human subjects and the measured values of S_vO_2 were in agreement with published data. Nevertheless, after more experience with this system, several limitations were noted involving the theoretical model, the image processing algorithms, and the system hardware. All of these concerns initiated the present work focusing mainly on improving the accuracy and the spatial resolution of the system.

The ratio-to-ratio approach was originally used for arterial pulse oximetry with the geometry of light transmission from a point source to a point detector. Whether this approach is still valid for the back reflection imaging system has not yet been demonstrated. Therefore, in the present work, we first used the Monte Carlo (MC) method to investigate the effectiveness of the ratio-to-ratio approach in calculating S_vO_2 for reflectance geometry with a planar wave illumination. Unlike the Beer–Lambert law or the hybrid model, the MC method accurately describes light propagation in a turbid medium (e.g., skin tissue) without any prior assumptions. If the ratio-to-ratio approach is verified as valid, then the quantitative relationship between the S_vO_2 and the ratio-to-ratio R needs to be identified. Different skin tissue models may also result in different relationships. To address this, we investigated the relationship for a semi-infinite and a two-layer skin model, and evaluated the variance in S_vO_2 when these two different models were used.

Regarding image processing, the surface of the imaged skin may not be flat, which results in a shape-based intensity bias. This surface modulation on the emitted light due to the curvature can induce error in the calculated S_vO_2. Therefore, curvature correction is necessary and should be included in the image processing algorithm. Also, since the venous blood volume alteration is induced by applying a pressure cuff, the cuff inflation may cause translational motion of the skin surface. The ratio-to-ratio approach is based on the change in emitted light intensity (or image pixel value) for the two different time points, e.g. before and after the venous occlusion. Surface motion causes pixel mismatch between the two images recorded at the two different time points, resulting in errors in the calculated S_vO_2 if the surface motion is not taken into account. When using a pixel binning or 2-dimensional moving average algorithm [19], the accuracy of the calculated S_vO_2 is acceptable for small motions but with deteriorated spatial resolution. To overcome this problem, we have now included a motion compensation algorithm in the image processing.

In addition to these improvements in modeling and image processing algorithms, we also improved the system hardware, including a ring light guide to guide the illuminating light perpendicularly to and uniformly on the skin surface, and cross-polarization detection to suppress surface specular reflection. With the upgraded system, we measured venous oxygen saturation in the palm, opisthenar (back of the hand) and wrist of human subjects. Compared with the original system, the spatial resolution in the S_vO_2 map was greatly improved, while the measured S_vO_2 values were comparable to previously published data.

2. Theoretical model and MC simulation

In order to obtain the S_vO_2 map from the collected images of the two wavelengths (i.e., 660 nm and 800 nm), the relationship between S_vO_2 and changes in the diffuse back reflectance must be established. Based on either the Beer–Lambert law [9] or the hybrid model [19], there is a linear relationship between S_vO_2 and the ratio-to-ratio R. However, neither the Beer–Lambert law nor the hybrid model is accurate enough for describing the diffuse back reflectance with a planar wave illumination on the skin. The Beer–Lambert law is suitable for the transmission geometry with a point source and a point detector. This is mostly because the difference in the path length for the two wavelengths is negligible in this measurement geometry. In the hybrid model, the path length difference is taken into account, and calculated with the light diffusion theory. However, the diffusion approximation is not accurate for describing the emitted light at each image pixel, because with a planar wave illumination, the path length distribution of the emitted light at each pixel contains abundant short paths comparable with the mean free path length. Therefore, an accurate relationship between S_vO_2 and R need to be investigated with a more accurate model.

MC simulation accurately describes light propagation in a turbid medium without any prior approximation [21,22]. In this section, we use Monte Carlo simulation to address two issues: first, with a certain S_vO_2, is the ratio-to-ratio R a constant over a wide range of venous volume alterations? If the answer is yes, which implies there is a one-to-one relationship between R and S_vO_2, then the second question is raised: what is the exact relationship between R and S_vO_2?

Unlike the light diffusion model, the Monte Carlo simulation treats the skin tissue as a turbid medium without any restriction on either optical parameters or geometric configurations. In our simulation, we used two types of skin models: one is a semi-infinite homogeneous model, and the other is a two-layer model. In the semi-infinite model we assumed the optical property is homogeneous. The two-layer model consisted of the epidermis and dermis layer. The epidermis layer was assumed to be 50 μm in thickness [10,23,24], containing melanin as the absorber, while the dermis layer was assumed to be semi-infinite, containing blood as the absorber. The optical parameters for these two models are listed in Table 1.

In the simulation, we assumed [10,23–25] the refractive index for skin tissue was approximately the same for the two wavelengths, $n=1.43$; the average hemoglobin concentration in the blood was 150 g/L; the volume fraction of the melanin in the epidermis was 2.0%; the volume fraction of blood in the dermis was 3%; and 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 were

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