Determination of human skin optical properties *in vivo* from reflectance spectroscopic measurements

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A novel approach has been proved to quickly and non-invasively determine the optical properties of human skin *in vivo*. It is based on the diffuse reflectance approximation model and subjected to the well established library of absorption spectra of water and hemoglobin. Under the nonlinear least-square algorithm, fitting the measured spectra in the range of 400—1000 nm to the diffusion approximation model, the reduced scattering coefficient and absorption coefficient of skin tissue can be quickly determined *in vivo*. The results show that this method is convenient and suitable for the real-time clinical application.

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The quick determination of optical properties of tissue *in vivo* is an important challenge in biomedical photonics. Several methods, including time-resolved, spatial-resolved, frequency domain and so on, have been developed in recent years[1−5]. Cheong *et al.*[6] reviewed the methods for measuring and calculating the scattering and absorption coefficients of tissue in detail. Some of these methods can non-invasively specify the absorption coefficient μ_a and the reduced scattering coefficient μ_s' *in vivo*, which require that the obtained experimental parameters should be sufficiently different so that they can separately specify the two unknown parameters (μ_a and μ_s'). For example, measurement of the escaping flux from a tissue at different distances from the source, often called steady-state spatial resolved diffuse reflectance $R(\rho)$, is sufficient to specify the optical properties at one wavelength of light. The disadvantage of steady-state spatial resolved diffuse reflectance is that it must measure at least eight different radial distances on the tissue surface, and the result is very susceptible to the random noise. The time-resolved reflectance technique can also non-invasively determine the optical tissue properties. However, the measurement system of this technique is very expensive[5]. Therefore, a method which can real-time determine the optical tissue properties *in vivo* is especially desired for the practical clinic application. Recently, Zhang *et al.*[7] presented a method to retrieve skin optical properties by total reflectance spectroscopy for the diagnosis of port wine stain. Johns *et al.*[8] investigated the reduced scattering coefficient of tissue phantom from reflectance spectroscopic measurement. Jacques[9] also pointed out the possibility to quickly determine the optical tissue properties using a fiber spectrometer. In this paper, a very convenient approach is given to quickly determine the skin optical properties *in vivo* using a simple optical fiber spectrophotometer. This method realizes the determination of the optical properties of living tissue, with the help of the library of absorption spectra for water and hemoglobin and a Mie-Rayleigh description of visible light scattering[9−12]. In addition, the method can overcome noise since it fits the data with known spectra over a wide band of wavelengths.

In this study, the optical properties of skin tissue, namely the reduced scattering coefficient μ_s' and the absorption coefficient μ_a can be quickly and non-invasively determined *in vivo*. This convenient approach is to measure the transport of light from a source fiber to a collection fiber, which are put in a pipe, slightly contacting the skin. The nonlinear least-square optimization method is used to analyze the measurement results. The experimental setup for reflectance spectroscopic measurement is shown in Fig. 1, which consists of a tungsten-halogen light source (LS-1, Ocean Optics, Inc.), focusing lens, fiber optic probe, spectrometer (USB2000, Ocean Optics, Inc.) with the spectral range of 400—1000 nm, and an IBM laptop computer. The optical fiber probe contains two 400-μm-diameter fibers for light delivery and collection respectively. White light is focused into an optical fiber, called the "source", which delivers the light to the skin surface as the fiber contacts slightly the skin. The second optical fiber, called the "collector", is placed at a distance of 0.18 mm to the source fiber. The collector fiber collects the diffused light from the skin, and brings them to the spectral device which measures the reflectance spectra range from 400 to 1000 nm. The

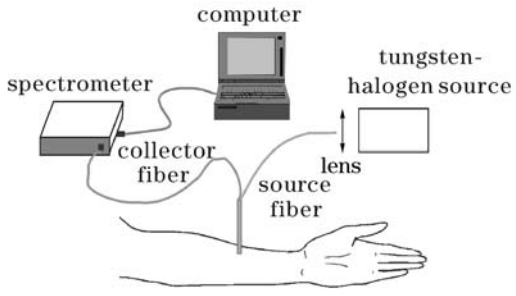

Fig. 1. Experimental setup for skin tissue reflectance spectroscopic measurement *in vivo*.

spectrometer is connected to the computer that acquires the spectra.

In order to model the reflected light from the skin tissue, a number of analytical expressions have been proposed[1,3,13,14]. Here we choose the Farrell model for a pencil beam irradiance[11]. In this model, the deeper dermis layer is assumed to be infinitely thick, implicitly assuming subcutaneous tissue, such as fat and muscle to be of negligible influence on the reflectance spectrum. This is a fair assumption for most visible wavelengths. In this regime, the spatial dependence of the diffuse reflectance can be described theoretically by

$$R_{\rm theo}(\rho) = \frac{1}{4\pi}\left[z_0\left(\mu_{\rm eff}+\frac{1}{r_1}\right)\frac{{\rm e}^{-\mu_{\rm eff}\cdot r_1}}{r_1^2} + (z_0+2z_{\rm b})\left(\mu_{\rm eff}+\frac{1}{r_2}\right)\frac{{\rm e}^{-\mu_{\rm eff}\cdot r_2}}{r_2^2}\right], \quad (1)$$

in which $r_1 = (z_0^2+\rho^2)^{1/2}$, $r_2 = [(z_0+2z_{\rm b})+\rho^2)]^{1/2}$. Herein, the extrapolated boundary for mismatched boundary condition is supposed, and $z_{\rm b} = 2AD$, A is an internal specular reflection parameter, $D = [3(\mu_{\rm a}+\mu_{\rm s}')]^{-1}$ is the diffusion constant, and $\mu_{\rm eff} = [3\mu_{\rm a}(\mu_s'+\mu_{\rm a})]^{-1}$ is the effective transport coefficient.

The reflection intensity from a standard reflectance is achieved by placing the source/collector probe at a certain distance from the standard reflectance. And then, the optical fiber contacts slightly with the skin to measure the reflection intensity from skin tissue, as shown in Fig. 2. Therefore, the measured reflectance $R_{\rm meas}(\rho)$ fitted by the reflectance model as

$$R_{\rm meas}(\rho) = G\cdot R_{\rm theo}(\rho) \quad (2)$$

can retrieve the relative parameters, namely the absorption and reduced scattering coefficients $\mu_{\rm a}$ and $\mu_{\rm s}'$. And the proportionality factor G in Eq. (2) contains factors like the optical fiber diameter, numerical aperture, and the coupling efficiency of the light into skin which is relatively constant with respect to wavelength[15].

To model the skin spectra and retrieve the optical properties of skin, the dependent wavelength parameters, the absorption and reduced scattering coefficients related to blood volume fraction and hemoglobin oxygen saturation should be determined. The epidermis with its keratin fibers appears to behave like dermis, so their reduced scattering coefficients can be approximated by that of dermis. Using the cylindrical Mie theory and the number density and average size of collagen fibers in skin, we can obtain

$$\mu_{\rm s}'(\lambda_{\rm Mie}) = 2\times10^5\times\lambda^{-1.5}. \quad (3)$$

While, skin tissue also exhibits a λ^{-4} like scattering in the Rayleigh limit of Mie scattering from structures much smaller than the wavelengths. Rayleigh scattering coefficient can be approximated as

$$\mu_{\rm s}'(\lambda_{\rm Rayleigh}) = 2\times10^{12}\times\lambda^{-4}. \quad (4)$$

The scattering behavior is dominated by Rayleigh scattering from small-scale structure at short wavelengths below 650 nm and by Mie scattering for wavelengths above 650 nm. Therefore, the visible to near-infrared (NIR) spectral region is significantly affected by both scatterings, that is

$$\mu_{\rm s}'(\lambda) = \mu_{\rm s}'(\lambda_{\rm Mie}) + \mu_{\rm s}'(\lambda_{\rm Rayleigh}). \quad (5)$$

It is well known that skin tissue represents a complex heterogeneous medium which can be divided into many layers according to its anatomical structure, such as stratum corneum, epidermis layer containing the melanosomes, dermis, and subcutaneous fat layers etc.. Therefore, the ultimate absorption coefficient of skin tissue contains contributions of all the absorbers in the given layers. However, since the stratum corneum and epidermis layers are much thinner than the others, in particular, the volume fraction of melanosomes in the epidermis is quite low, about 1.3%—6.3% for lighted-skin adults[9], for simplicity, we can ignore their contributions to the absorption in the visible and NIR range. This has been well established by Jacques[9]. Therefore, we assume that the oxy-/deoxy-hemoglobin and water dominate the absorption of the reflectance spectra in this study, namely

$$\mu_{\rm a}(\lambda) = B\cdot[S\cdot\mu_{\rm a-oxy} + (1-S)\cdot\mu_{\rm a-deoxy}] + W\cdot\mu_{\rm a-water}, \quad (6)$$

where B is the blood volume fraction, S is the hemoglobin oxygen saturation. And then, the optical properties of skin tissue can be determined by using Eqs. (5) and (6) in the following three steps. Firstly, build a database of the absorption spectra for water, oxy-, and deoxy-hemoglobin according to the present references[9−11], as shown in Fig. 3. Secondly, keep the water content of skin

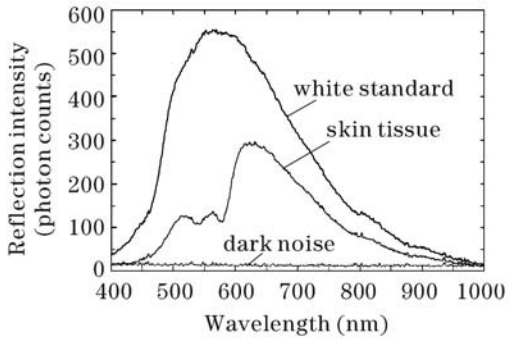

Fig. 2. Reflection intensity of skin tissue and the standard reference versus wavelength.

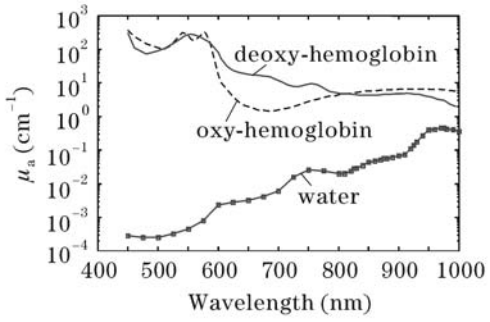

Fig. 3. Absorption coefficients $\mu_{\rm a}$ of oxy-, deoxy-hemoglobin and water versus wavelength.

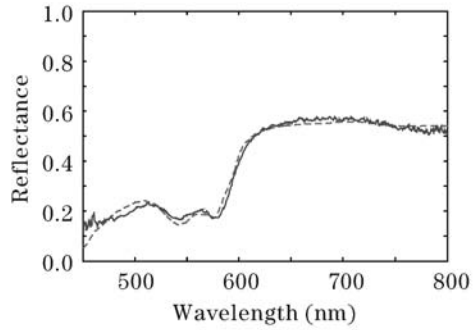

Fig. 4. Measured (solid curve) and fitted (dashed curve) spectra of the skin tissue *in vivo*.

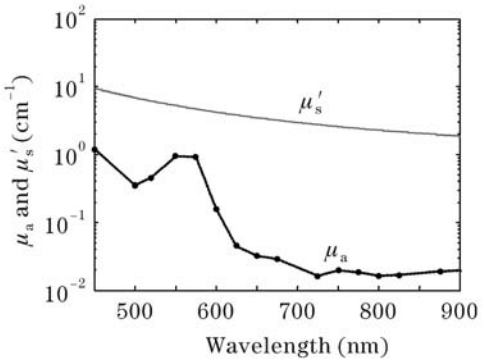

Fig. 5. Spectra of the absorption coefficient μ_a and reduced scattering coefficient μ_s' of the skin tissue.

Table 1. Results of Forearm Skin Measurement *in vivo* at 800 nm

	B (%)	S (%)	μ_a (cm^{-1})	μ_s' (cm^{-1})
This Work	0.35	54	0.16	2.3
Ref. [16]	2.2	74	0.65	8.0
Ref. [17]	–	64	0.19	6.0

tissue (W) and tissue index (n) constant in this study ($W = 0.70$, $n = 1.4$). Thirdly, suppose the initial values of the relative parameters B, S, and G to start the fitting. The initial values are set as $B = 1.0\%$, $S = 60\%$, and $G = 0.5$ according to Refs. [6,7,9,15]. The measured and fitted spectra are shown in Fig. 4.

From Fig. 4, we can see that the measured spectra of the forearm in the wavelength range of 400—800 nm agree very well with the fitted spectra. The spectra of the obtained absorption and reduced scattering coefficients of skin tissue as functions of wavelength are shown in Fig. 5. We also compared the results of this work with those of Refs. [16,17], as shown in Table 1. One can see from the table that our results agree better with those of Matcher's[17] than Doornbos'[16]. The overestimation of the absorption coefficient in Ref. [16] may result from its higher tissue blood percentage, which has a reasonably physiological value ranging from 0.2% to 1% for skin tissue[3]. And the discrepancy of μ_s' may be because of the difference of individuals, including the skin color and the age.

In conclusion, this study has shown that the optical properties of skin tissue can be determined quickly *in vivo* by combining reflectance spectrum measurement with a library of absorption spectra of water and hemoglobin. By fitting the experimental data under the nonlinear least-square algorithm to the diffusion approximation model, important physiological parameters, such as the blood volume fraction and oxygen saturation, have been retrieved, which in turn verifying the effectiveness of the proposed method. This method is proved to be able to determine the optical properties of living tissue *in vivo* quickly and non-invasively. It is promising in clinical applications, such as skin diseases diagnosis and the real-time measurement of optical tissues for therapy plan.

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