Optik 126 (2015) 460–463

Contents lists available at ScienceDirect

Optik

journal homepage: www.elsevier.de/ijleo

Noninvasive measurement of blood hemoglobin concentration by time-resolved transmission spectroscopy

M.X. Sun [a], L.Y. Zhang [b], C.Y. Jiang [a], Y. Yuan [a], J.C. Yang [a], L. Tian [a], J.P. Wu [a], Y.X. Li [a,*]

[a] Laser Medicine Laboratory, Institute of Biomedical Engineering, Chinese Academy of Medical Science & Peking Union Medical College, Tianjin 300192, China
[b] Basic Department of Military Transportation College, Tianjin 300161, China

ARTICLE INFO

Article history:
Received 14 December 2013
Accepted 20 July 2014

Keywords:
Blood hemoglobin concentration
Time-resolved transmission spectroscopy
Noninvasive measurement
Laplace transform

ABSTRACT

An optical spectroscopic method is investigated theoretically for noninvasive and continuous monitoring of blood hemoglobin levels. The method may provide accurate measurement of hemoglobin concentration by detection and analysis of dynamic dual wavelength time-resolved transmission under a condition of artificial blood flow kinetics in a human finger. This condition can assist light scattering and leads to a better signal-to-noise ratio necessary to accurate measurement of transmission. The modified parametric slopes were derived from the Laplace transformed data of the time-resolved transmittance. Continuous hemoglobin concentration was measured in the range from 6 to 16 g/dl and the results show that the sensitivity of hemoglobin sensing is enhanced when we use selected values of the Laplace transform parameter $p = 5 \times 10^{10}\ s^{-1}$ comparing with the continuous wave situation.

1. Introduction

Hemoglobin (Hb) is arguably the most important component in our blood, and is responsible for transporting oxygen from the lungs to the rest of our body. The blood hemoglobin concentration is among the most commonly performed of all clinical laboratory tests. Hemoglobin levels in humans range from 12 to 16 g/dl [1]. Hemoglobin measurements allow the detection of anemia and hemorrhage, and are widely used by hospital wards such as operating rooms, ICU, emergency and delivery rooms as well as in outpatient clinics and women's and children's health care facilities.[2] In blood donation centers, a predonation Hb measurement of blood donors is required by most countries to protect anemic donors, and to ensure adequate Hb content of blood units.

Currently, only two methods, both invasive, are available for Hb measurement [3]. The first, most common method requires withdrawal of a blood sample, which is then transported to and analyzed in a clinical laboratory. Blood sampling is invasive, discontinuous, labor intensive, and aggravates perioperative anemia. In addition, clinical decisions must await return of results from a laboratory. The second method, used only during cardiopulmonary bypass surgery or hemodialysis, optically monitors the Hb in extracorporeal blood circuits. Consequently, a noninvasive, continuous monitor of the Hb would be valuable.

In earlier studies, we proposed to use a time-resolved approach combined with occlusion spectroscopy (TRACOS) for in vivo noninvasive measurement of blood glucose concentration [4]. This method is based on dynamic dual wavelength time-resolved measurements under a condition of artificial blood flow kinetics in a human finger. The simulation results show that the sensitivity of glucose sensing is enhanced when late arriving photons are emphasized by Laplace transforming the time-resolved transmittance with a small negative parameter. In this study we present theoretical investigations on an in vivo hemoglobin measurement by TRACOS. Modified parametric slopes are derived from Laplace transformed time-domain data. Numerical simulations have been carried out to correlate the modified parametric slopes to the blood hemoglobin concentration. Continuous hemoglobin concentration was measured in the range from 6 to 16 g/dl and the results show that the sensitivity of hemoglobin sensing is enhanced when we use selected values of the Laplace transform parameter $p = 5 \times 10^{10}\ s^{-1}$ comparing with the continuous wave situation.

2. Modeling of absorption and reduced scattering coefficients associated with hemoglobin

The hemoglobin concentration affected the absorption and reduced scattering coefficients of the blood. In occlusion

* Corresponding author.
E-mail address: yingxinli2005@126.com (Y.X. Li).

http://dx.doi.org/10.1016/j.ijleo.2014.07.148
0030-4026/

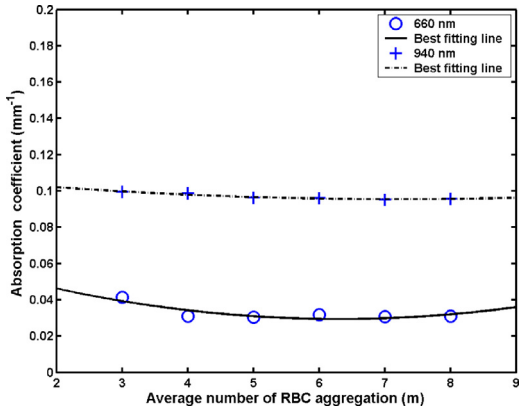

Fig. 1. Blood absorption coefficients as a function of average number of RBC aggregation.

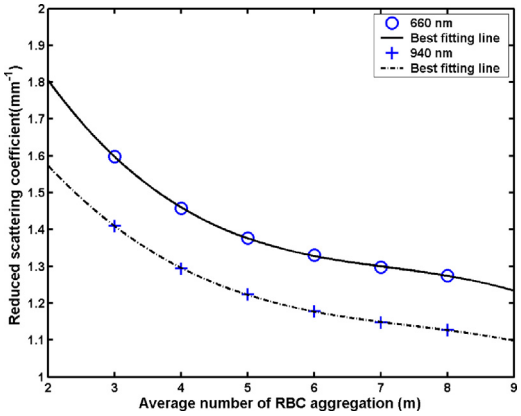

Fig. 2. Blood reduced scattering coefficients as a function of average number of RBC aggregation.

spectroscopy, a temporary over-systolic pressure was imposed on a finger to stop the blood flow and diminish the shear force. In low flow rates, red blood cells (RBCs) aggregate along their axis of symmetry and form so-called rouleaux [5]. The increase in the size of scattering particles (RBC aggregation) causes the changes of the scattering coefficient, and thus changes the light transmittance. The method for simulating optical properties under this experimental condition, previously described in detail [4], the aggregation is thought to be one-dimensional, i.e. erythrocytes aggregate side by side. According to this model, the aggregate shape is approximated by spheroid. RBC and RBC aggregations are modeled as random spheroids. Their scattering cross section, absorption cross section and anisotropy are computed by using the T-matrix method.

The blood absorption coefficient and scattering coefficient are related to the absorption and scattering cross-sections of individual RBC aggregates and their concentration. The absorption and scattering coefficients are given by [6]:

$$\mu_a = \rho\sigma_a = \frac{H\sigma_a}{V_0}$$
$$\mu_s = \rho\sigma_s = \frac{H\sigma_s}{V_0} \quad (H < 0.2) \tag{1}$$
$$\mu_s = \rho\sigma_s = \frac{H(1-H)\sigma_s}{V_0} \quad (H \geq 0.2)$$

Here σ_a and σ_s are absorption and scattering cross-sections of an individual RBC, μ_a and μ_s are blood absorption and scattering coefficients. The above equations can be generalized to RBC aggregates. In case of RBC aggregation, H remains constant while V_0 becomes the volume of a RBC aggregate and the cross-sections are of individual aggregates. Hemotcrit is strongly correlated with hemoglobin concentration. An estimated hematocrit as a percentage may be derived by tripling the hemoglobin concentration in g/dl and dropping the units [7]. The blood absorption and scattering coefficients as functions of average number of RBC aggregation are shown in Figs. 1 and 2.

From the figures, the increase in the average size of the scattering particles caused by RBC aggregation can induce changes in the optical scattering characteristics of blood dramatically during the stage of blood flow cessation. These dynamic changes can be readily measured using spectroscopic methods and differentiated from background optical properties of other tissues.

3. Method of hemoglobin determination

The hemoglobin concentration affects the imaginary part of refractive index of the RBC expressed by Eq. (2) and therefore has an influence upon the absorption properties of particles (RBC) suspended in blood.

The cell with the imaginary part given by [8]

$$n_i = \frac{\ln 10}{4\pi}\frac{[c]\lambda_0}{n_r}(h_1\gamma_1 + h_2\gamma_2), \tag{2}$$

where λ_0 is the light wavelength in free space, $[c]$ is the hemoglobin concentration. h_1, h_2 $(=1-h_1)$, γ_1, and γ_2 are the volume ratios, molar extinction coefficients of HbO_2 and Hb, respectively. Hemotcrit has an effect on the determination of absorption and scattering coefficients from Eq. (1). Hemotcrit is strongly correlated with hemoglobin concentration. An estimated hematocrit as a percentage may be derived by tripling the hemoglobin concentration in g/dl and dropping the units.

A 10 mm thickness finger is modeled as a single-layer sample. We consider a simplified medium consisting of blood plasma and RBC aggregations inside instead of a very complicated real medium containing also tissue, skin, bones, etc. Light sources and detectors operating in the red/near-infrared (RNIR) spectral region and pneumatic cuffs that produce over-systolic pressure to occlude blood flow are included in the noninvasive blood hemoglobin probe. An over-systolic pressure is imposed on finger to make blood flow cession and induce scattering change due to RBC aggregation. Two laser beams of wavelengths of 660 nm and 940 nm, respectively are normally incident on the sample surface. A time-resolved optical detector is used to receive the transmittance at the opposite position of source. The time-resolved transmittance is given by the following formula based on the diffusion equation [9],

$$T(l,t) = (4\pi Dc)^{-3/2}t^{-5/2}\exp(-\mu_a ct)\times\left\{(l-z_0)\exp\left[-\frac{(l-z_0)^2}{4Dct}\right]\right.$$
$$-(l+z_0)\exp\left[-\frac{(l+z_0)^2}{4Dct}\right]+(3l-z_0)\exp\left[-\frac{(3l-z_0)^2}{4Dct}\right]$$
$$\left.-(3l+z_0)\exp\left[-\frac{(3l+z_0)^2}{4Dct}\right]\right\} \tag{3}$$

where $D=\{3[\mu_a+(1-g)\mu_s]\}^{-1}$ is the diffusion coefficient, g is the assembled mean of cosine of scattering angle. Here $z_0=[(1-g)\mu_s]^{-1}$ is the depth of one transport mean free path in the medium, c is the speed of light in the tissue, l is sample thickness and it is 10 mm in our simulation.

Parametric slope is defined as a ratio between a first function depending on a light response of the medium corresponding to one wavelength and a second function depending on the light response of the medium corresponding to another wavelength. In this paper, we defined a modified parametric slope MPS, which is derived