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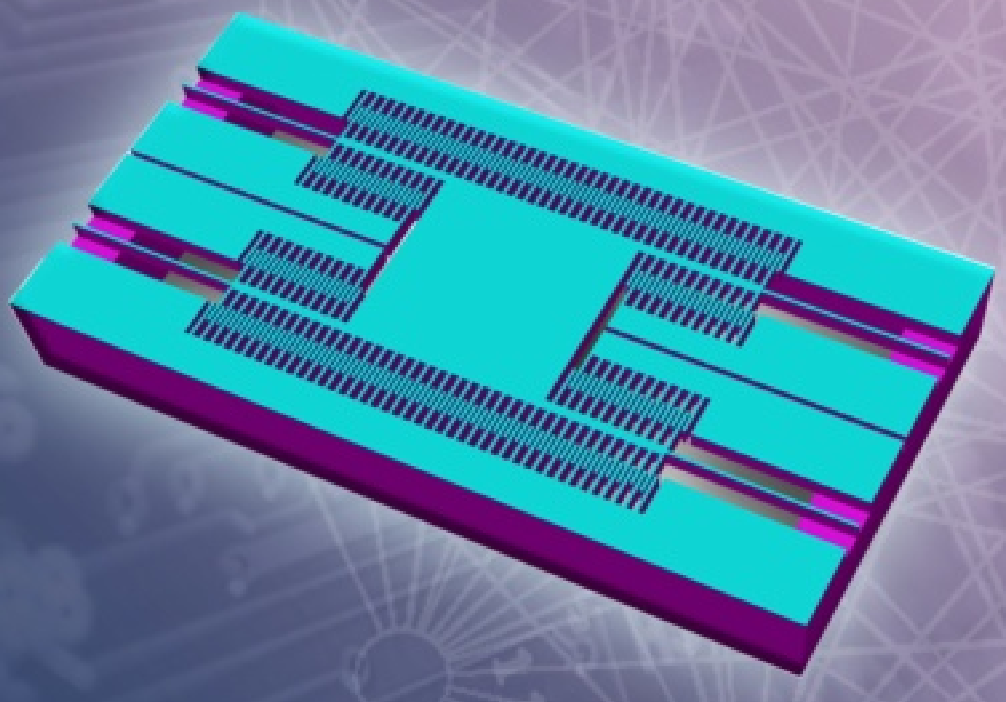

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Extracting Broadband Tissue Optics Parameters from One Source-Detector CW Diffuse Optical Spectroscopy

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Abstract: Human tissue by its nature is a kind of complex entity in structures and compositions, with their respective spectral properties. These properties make complex light-tissue interactions as the light propagates inside the tissue. Hence the information on broadband wavelength-dependency of optical parameters is important to asses the state of investigated tissue. Accurately determined tissue optics parameters (i.e. the absorption and scattering coefficients) are crucial for further derivation of many other ones related to hemodynamics states. Nowadays the Diffuse Optical Spectroscopy, which is based on spectral measurements of diffusely re-emanated of injected light source, plays an important role in many tissue diagnostics applications. Most of available extraction schemes are based on multiprobe/multidistance spectral measurements. This paper describes a novel and simple extraction scheme from only single distance of source-detector probe. The use of single pair source-detector probe will significantly reduce the complexity and cost of measurement system.

Keywords: Broadband tissue optical parameters, Diffuse optical spectroscopy, Tissue-parameters extraction-scheme, Single pair source-detector probe.

1. Introduction

Recently the Diffuse Optical Spectroscopy (DOS) techniques have been proven to play a significant role for many diagnostics applications. These techniques are based on the characteristics of diffuse photon transport in tissue, which can be used to investigate tissue's physiological states up to several centimeters below the tissue surface [1]. Such advantageous benefits are due to availability of a *spectral window* in the range of 600 – 900 nm, in which the photon transport inside tissue are dominated by scatterings rather than absorption events. This optical window was first accidentally discovered by Jöbsis in 1976 [2].

Accurate extraction of tissue's absorption- and reduced scattering coefficients from the detected diffusely optical signals is crucial for many diagnostic and therapeutic applications. These optical properties are sensitive to any changes in tissue's micro-structures (i.e. collagen fibers, nuclei and other particulates), as well as to changes in chromophores due to disease's induced abnormal-conditions. Development of optical quantification techniques, i.e. the DOS techniques, is intended to be capable to exploit these changes into what so called *diagnostically-valued information*. Taking advantage of these optical properties changes, accurate differentiation the state of tissue, which reflect the physiological state of the investigated tissues, can be reached even in their earliest phase [3, 4].

Based on the types of optical source used and the way how the photon being detected, there are three modes of system that are normally used in Diffuse Optical Spectroscopy. They are the *continuous wave* system, the *frequency-domain* system, and the *time-domain* system. The continuous wave system is regarded as the simplest, and it enables fast data-acquisition with the use of simple as well as low-cost detectors and related electronics detection system. Meanwhile the rest two modes utilize source's intensity-modulation system to produce a modulated diffusing wave. The information content per measurement from the last two modes show an increasing tendency, as a trade-off for their complexities and cost of the systems. Brief descriptions among these modes of detection are given in [5].

The complexities of actual human tissue are commonly being modelled as complex-layered turbid-media with their own specific biophysical and biochemical states. Both states are spectrally dependent in nature. The tissue's scattering properties are more related to morphology, while their absorption properties are related to biochemical states. So accurate determination of tissue optical properties, which relevant to asses respective physiological states will require multiple-wavelengths measurements. Several efforts have been developed to separate these interfering properties, among other are the ones using multiple single-wavelength techniques: either in continuous-wave mode [6] or frequency-domain mode [7]. Approach to implement broadband wavelength-range measurement will benefit to simultaneously measure the concentrations of several absorbers with improved accuracy, and at the same time also get a better insight into spectral dependence of the accompanying scatterings [8, 9].

The work described in this paper is based on the continuous-wave mode of diffuse optical spectroscopy. Under this technique, working principle of the system is as follow: a continuous pencil beam of light is injected into the tissue, and the diffusely re-emanated beams (i.e. the beam that have already experienced multiple scattering events by tissue's scatterers and multiple absorption events by tissue's chromophores) are coupled through collecting fibers. These collecting fibers are located at several distances from the source for further detection by spectrometer. This technique is known as *spatially-resolved reflectance measurements*. Calibrating the inherent responses of the system by using standard integrating sphere measurement in addition to dark currents substraction, the tissue spectral reflectance data can be obtained.

Tissue optical properties extracted from the measured diffuse reflectance data are then obtained by fitting these measured data with the theoretical diffuse reflectance equations that are derived from the diffuse approximation of light-transport equation inside the tissue. Most previous works in this field referred to the approach, which were originally proposed by Farrel & Patterson [10] in deriving the equation. Separating the coupled absorption- and scattering components can be achieved by simultaneously fitting multiple source-detector measurements [9, 11-14].

In this paper an extraction scheme for broadband tissue optical parameters from diffuse measurements using only single pair of source-detector probe is described. This proposed scheme utilizes a tissue model with respected *a priori* spectral form of chromophore components for the calculation of total absorption coefficient. Together with the Mie scattering model, these coefficients are substituted into the forward-calculation of theoretical diffuse reflectance equation. Using a Lavenberg-Marquadt based nonlinear fitting algorithm, the related parameters from a tissue model can then being extracted. These parameters will be used to derive other parameters related to tissue hemodynamic states, i.e. the broadband absorption and reduced-scattering coefficients, the tissue oxygen saturation (StO_2), and the total hemoglobin concentration. Results of derived parameters are then being compared with ones which are multi-spectrally measured using the ISS Oxymeter (a frequency-domain DOS system).

2. Theories

2.1. Radiative Transfer Equation and Diffuse Approximation

Injection of light beam to probe the tissue will be followed by some consecutive light-tissue interactions events. These interactions are preceded with reflections at tissue boundaries, and then continued with absorption due to tissue chromophores and fluorophores components, as well as by multiple scattering events due to tissue scatterers. All of these interactions will finally lead to the attenuation of intensity as the light re-emanated from the tissue for further detection.

The tissue-light interactions are commonly being modelled by using what so called as *Radiative Transfer Equation (RTE)*. This mathematical model can be described based on schematical sketch in Fig. 1.

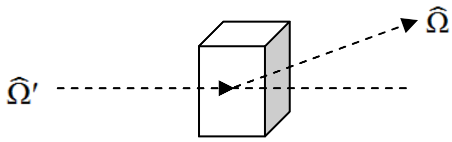


Fig. 1. Radiance of light at position *r* travelling in a direction of $\hat{\Omega}$, which is decreased by absorptions and scatterings but also increased by light that are scatterred from any of $\hat{\Omega}'$ into direction of $\hat{\Omega}$.

As shown in Fig. 1, the light transport at a position inside the tissue can be formulated using the following equation:

$$\frac{1}{v}\frac{\partial L(r,\hat{\Omega},t)}{\partial t} + \nabla . L(r,\hat{\Omega},t)\hat{\Omega} + \mu_t L(r,\hat{\Omega},t) = \mu_s \int L(r,\hat{\Omega}',t) f(\hat{\Omega},\hat{\Omega}') d\hat{\Omega}' + S(r,\hat{\Omega},t) \quad (1)$$

In the left-side of equation, the first, second, and third terms respectively refer to the time-derivative of radiance (number of photons entering element minus ones that leaving), flux of photons along direction of $\hat{\Omega}$, and absorption and scattering within the element. While the first and second terms in the right-side equation refer to the integral accounts for photons at position of *r*, which are contributed from scattering from all directions of $\hat{\Omega}'$ into $\hat{\Omega}$, and the source of photons. The parameter μ_t is the transport coefficient (which is equal to $\mu_a+\mu_s$, i.e. the absorption and scattering coefficients, respectively) and the $f(\hat{\Omega},\hat{\Omega}')$ is known as the normalized *phase function* (represents the probability of scattering from direction of $\hat{\Omega}'$ into $\hat{\Omega}$).

In the “spectral window” region (600 – 900 nm) the multiple scattering events will predominate the absorption events (i.e. 10 ~ 1000 times greater [15]). Under this condition the light transport in tissue can be regarded as *diffusely propagated*. And thence the diffuse equation is valid to approximate the radiactive transport equation of light in tissue. The equation (1) can then being written as:

$$\frac{\partial}{\partial t}\Phi(\boldsymbol{r},\mathrm{t}) = \nabla . D\nabla\Phi(\boldsymbol{r},\mathrm{t}) - v\mu_a\Phi(\boldsymbol{r},\mathrm{t}) + v\mathrm{S}(\boldsymbol{r},\mathrm{t}) \quad (2)$$

where $\Phi(r, \mathrm{t})$ and υ are the photon fluence rate and speed of light in the tissue, respectively. The term $D = 1/[3(\mu_a+\mu'_S)]$ is the photon diffusion coefficient, with μ_a and are the absorption and reduced scattering coefficients, respectively. Meanwhile the term S represents the isotropic source (which gives the number of photons emitted at position r). The tissue model can then being regarded as a homogeneous semi-infinite media, where both of source and detector measuring fibers reside on the same tissue’s surface and with separation distance (ρ) that should be much higher than the transport’s mean free-path (which is equal to $1/\mu_t$).

Solution to the diffuse reflectance $R[\rho,\mu_a(\lambda),\mu_S'(\lambda)]$ from semi-infinite media with a continuous-wave of light excitation can be derived by applying the extrapolated boundary condition [11, 16], and can be written as:

$$R[\rho,\mu_a(\lambda),\mu_s'(\lambda)] = C_1\,\Phi(\rho) + C_2\,j_z(\rho) \quad (3)$$

$$\Phi(\rho) = \frac{1}{4\pi D}\left[\frac{exp\left(-\mu_{eff}r_1(\rho)\right)}{r_1(\rho)} - \frac{exp\left(-\mu_{eff}r_2(\rho)\right)}{r_2(\rho)}\right] \quad (4)$$

$$j_z(\rho) = \frac{1}{4\pi}\left[\frac{1}{\mu_t}\left(\mu_{eff} + \frac{1}{r_1(\rho)}\right)\frac{exp\left(-\mu_{eff}r_1(\rho)\right)}{r_1{}^2(\rho)} + \left(\frac{1}{\mu_t} + 2z_b\right)\left(\mu_{eff} + \frac{1}{r_2(\rho)}\right)\frac{exp\left(-\mu_{eff}r_2\right)}{r_2{}^2(\rho)}\right] \quad (5)$$

with

$$\mu_{eff} = \sqrt{3\mu_a(\mu_a + \mu_s')} \qquad z_b = \frac{2A}{3\mu_t} \quad (6)$$

The term A is defined as the *internal reflection coefficient*, and its typical value for most tissue is approximately 3.23 (for a refractive index mismatch of 1.4 [10, 11, 17]). Meanwhile $r_1(\rho)$ and $r_2(\rho)$ can be expressed as:

$$r_1(\rho) = \sqrt{\left(\left(\frac{1}{\mu_t}\right)^2 + \rho^2\right)} \qquad r_2(\rho) = \sqrt{\left(\left(\frac{1}{\mu_t} + \frac{4A}{3\mu_t}\right)^2 + \rho^2\right)} \quad (7)$$

For large source-detector distance, i.e. $\rho >> \frac{1}{\mu_t}$, then $r_1 = r_2 \approx \rho$ and the reflectance equation (3) can be simplified into the following equation:

$$R[\rho;\, \mu_a(\lambda),\mu_s'(\lambda)] = \frac{0.62}{(\mu_a+\mu_s')}\left(\sqrt{3\mu_a(\mu_a + \mu_s')} + \frac{1}{\rho}\right)\frac{exp\left(-\sqrt{3\mu_a(\mu_a+\mu_s')}r\right)}{\rho^2} \quad (8)$$

2.2. Spectral Form of Chromophore Components

The main cromophores components of tissue are the oxygenated hemoglobin (HbO_2), deoxygenated hemoglobin (Hb), water, and fat. In addition, there are other chromophores with minor availability and absorption effects [18]. All of these chromophores are wavelength-dependent in nature, and will contribute to the total absorption of the transported light in tissue. The spectral behavior of these four chromophores can be shown in Fig. 2.

This spectral behavior of the chromophore components will be incorporated as a priori information for calculation of the total absorption coefficient using theoretical diffuse reflectance. A tissue model that represents the types of chromophores and scatterers will be used for fitting the reflectance data, as will be described in next section.

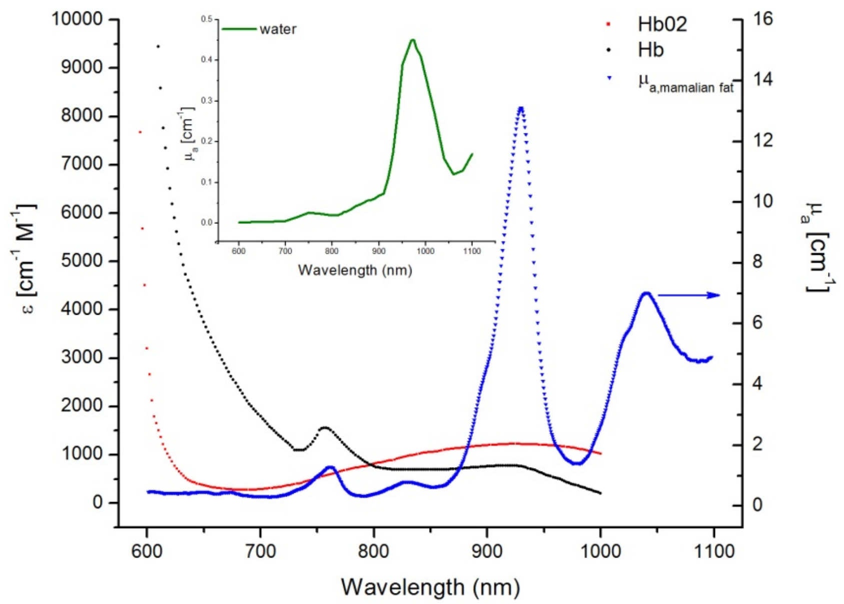


Fig. 2. Spectral behavior of some main tissue chromophores.

2.3. Tissue Model

A tissue model is defined as a model for absorption and scattering characteristics of the tissue due to its constructing components. The model can be set as the simplest one up to the more complex ones, which are more closely mimic the actual conditions of the investigated tissue. Model of absorption coefficient used in this work is adopted from the work of Nachabé et al. Missing a component to incorporate into the tissue model will reduce the accuracy of the extraction process [19]. The used model can be given in the following equation for absorption and scattering.

$$\mu_a(\lambda) = 2.303\left(C_{H_bO_2}\epsilon_{H_bO_2} + C_{H_b}\epsilon_{H_b}\right) + f_w\,\mu_{water}(\lambda) + f_{Fat}\,\mu_{Fat}(\lambda) \tag{9}$$

$$\mu'_s(\lambda) = A\lambda^{-B} \tag{10}$$

The $C_{H_bO_2}$ and C_{H_b} terms are concentrations of oxygenated and deoxygenated hemoglobin, respectively, and $\in_{H_bO_2}$ and $\in_{H_b}$ are the respective spectral extiction coefficients [cm^{-1} M^{-1}]. Meanwhile the f_w and f_{Fat} terms represent the volume fraction of water and fat in the in investigated volume. The scattering component of the reflectance is modelled using Mie's scattering equation, with the A and B are coefficients which represent scattering amplitude and scattering power, respectively. The use of Mie scattering in the model is due to predomination of this scattering regime in the spectral region of interest, in compare to the Rayleigh one.

3. Materials and Methods

3.1. Measurement System

For acquiring diffuse reflectance data, measurements were conducted using the Jaz, a modular white-light mini-spectrometer from Ocean Optics. This spectrometer is equipped with the diffraction grating with detection range of 500 – 1100 nm. Two patch-cord optical fibers (Vis-NIR) are connected to the spectrometer and LS-1 Tungsten Halogen Light Source (360 – 2500 nm), meanwhile their other ends are connected to measuring probe. The probe is constructed from two SMA 905 half-sleeve (11 mm) which are inserted into a black elastic rubber pad with separation distance of 1 cm.

Calibration of the intrinsic response of the measurement system was done by detaching the both ends of patch cord fibers and connecting to the FOIS-1 Fiber Optic Integrating Sphere, which has a spectralon-coated sphere inside with a high and flat reflectance in the spetral region of 200 – 2500 nm. All of the acquisition sequence is controlled with the *Spectra Suite* software.

3.2. Algorithm for Parameters Extraction

As briefly described in Section 1, the related tissue parameters can be extracted from the measured reflectance data through a fitting process. A Lavenberg-Marquadt based nonlinear curve-fitting process is adopted to fit the measured reflectance data with the theoretical forward calculation, which is derived from the diffuse equation (8) and by incorporating tissue model as given in equation (9-10) with their respective *a priori* spectral form. The implemented process can be schematically depicted in Fig. 3. The process of loading and preprocessing of the acquired reflectance data, as well as nonlinear fitting sequence, were implemented using scripts on Origin Pro 8 (from Origin Lab Corporation).

3.3. Tissue Measurements and Benchmarking using the ISS (Frequency Domain) Oxymeter

Measurement tests on tissue were accomplished on the forearm at positions as indicated in the Fig. 4 below. Acquisition time for single broadband scan was set as 4.1 ms, and averaging of 10 scans were used to increase the SNR of the reflectance data.

These reflectance measurements are then being compared with measurements at similar locations done using the Imagent frequency domain oxymeter (from ISS), which was operated under wavelengths of 690 nm and 830 nm.

3.4. Preliminary Calibration using Intralipid Liquid Phantom

In order to improve the accuracy of parameters extraction process, a standard calibration procedure is normally required. One of the commonly adopted calibration standard is done by using Intralipid solution as a tissue simulating phantom. Intralipid solution is a kind of pharmaceutical product, made of fat droplets from soya bean oil and glycerol which are suspended in water. This solution is normally used for parenteral nutrition for patients. In the market, the Intralipid solution is available for three different fat concentrations, i.e. 10 %, 20 %, and 30 %. This standard solutions show a high stability and small batch-to-batch variations. Preparing tissue phantom with certain optical properties (reduced scattering coefficient) can be achieved by dilution the solution with distilled water until certain intended concentration achieved [20].

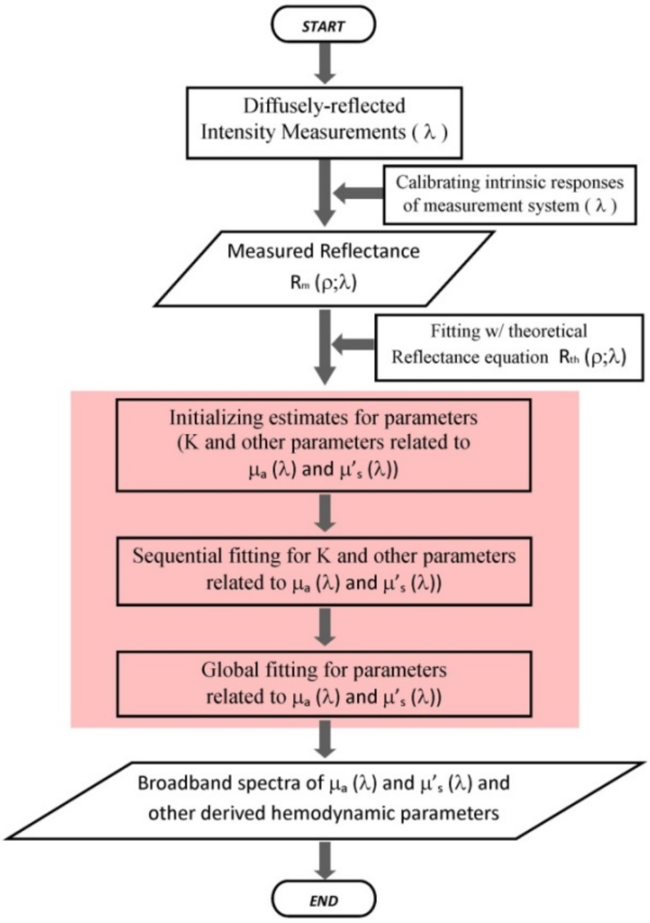


Fig. 3. Algorithm for Parameters Extraction.

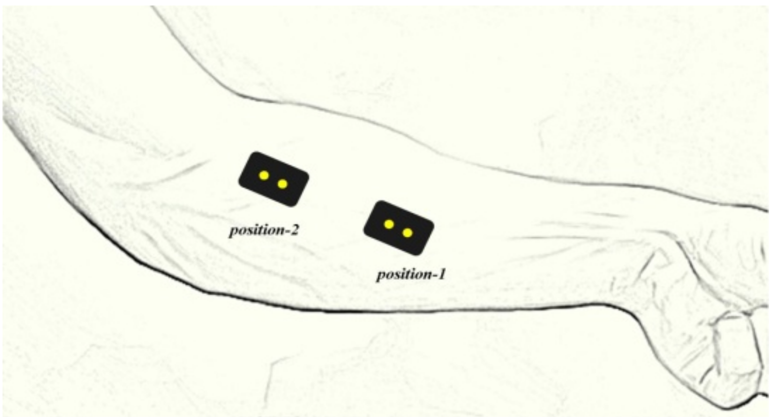


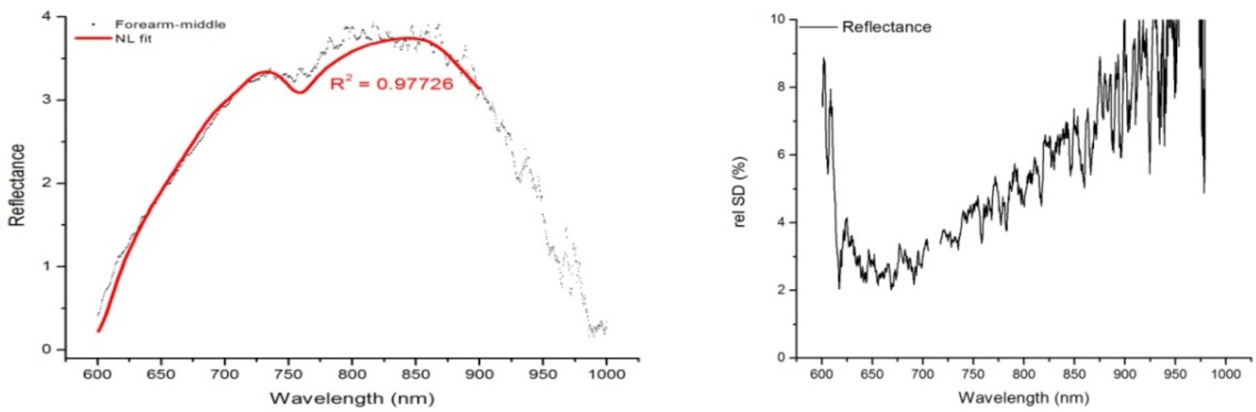
Fig. 4. Measurement Positions on Forearm.

To simulate variation in tissue absorbing properties, additional drops of India ink can be used. India ink is a kind of particulate absorber, i.e. insoluble carbon particles which are suspended in aqueous medium. There are plus and minus in using India ink as simulating absorber. Chemically as well as spectroscopically stable, it does not fluoresce, and its non-toxicity are the plus factors, while the minus one is due to its suspension state. Thence it will not only absorb, but it will also scatter the light [21].

Preparation of the liquid tissue phantom follows procedures given in [22] to reach desired value of reduced scattering coefficient (μ'_S) of 8 cm^{-1} (@ 830 nm). A volume of 195.26 mL Intralipid 30 % is diluted to make a total volume of seven liter. The dilution is prepared in a fish tank, and is stirred evenly with a mixer to homogenize it. Additional drop of 0.98 mL India ink (10 % vol) is given to achieve desired absorption value of 0.05 cm^{-1} (@ 830 nm). Subsequent additional 1.29 mL drops of India ink (10 % vol) are added to make incremental absorption values of 0.025 cm^{-1} (@ 830 nm). Both preparation and extraction of parameters are done by incorporating absorption from water, in addition to the one of the India ink, and their related scattering factor. Diffuse reflectance data are then acquired for each previously described values, and calibrating the extracted parameters from measurements using prescribed values from prepared standard solutions.

3. Results and Discussion

The typical spectral reflectance data obtained from measurement (together with nonlinear fitting result using the proposed scheme) can be seen in Fig. 5a, meanwhile the measurement precision in the spectral region of interest can be seen in Fig. 5b. Decreasing in precision level of the spectral reflectance data can be understood as the decreasing sensitivity of the spectrometer's detector, particularly at region near both ends of its spectral detectability.

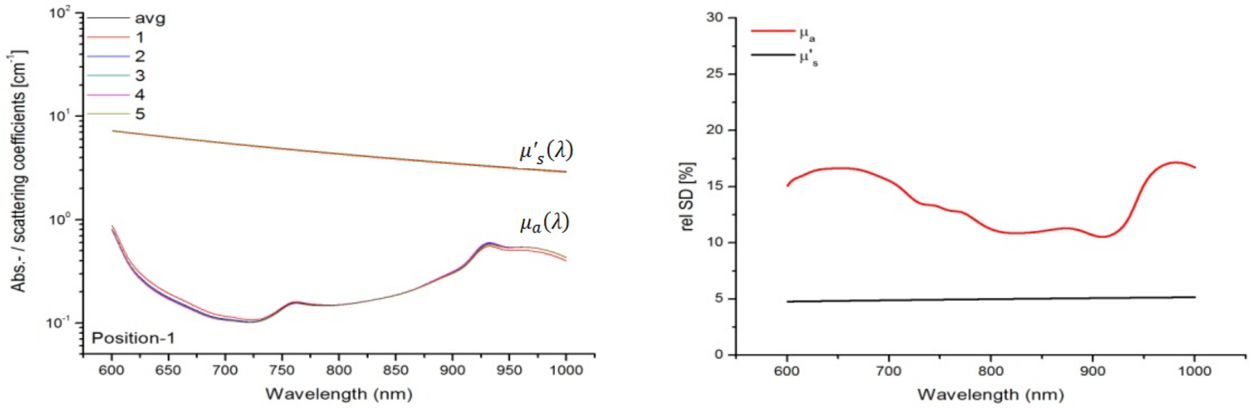


(a) (b)

Fig. 5. (a) Typical spectral reflectance data from measurement and corresponding nonlinear fit using proposed scheme, (b) Precision level of reflectance data.

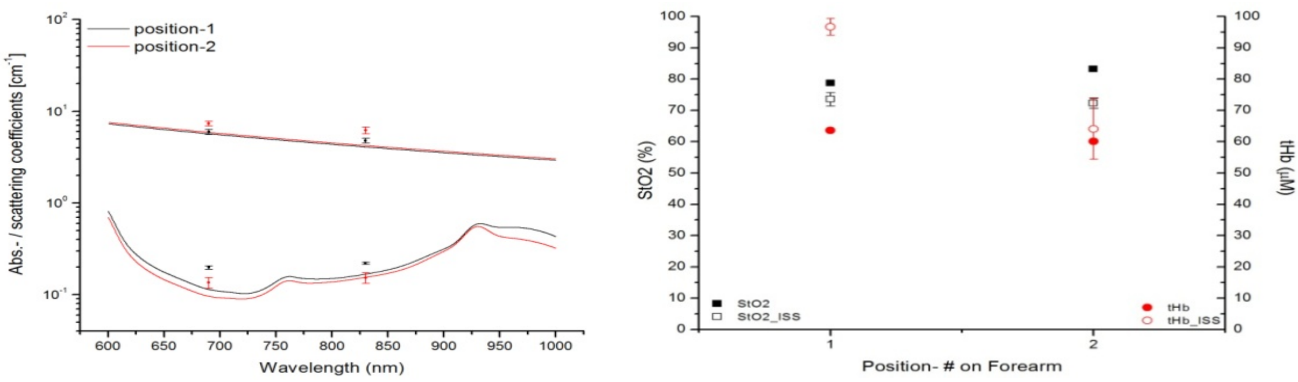
The values of broadband absorption coefficient $\mu_a(\lambda)$ and the reduced scattering coefficient $\mu'_S(\lambda)$ are given in Fig. 6a. The repeatability of the parameters extraction was also assessed using five consecutive reflectance measurements. These results are in the range of 10 – 15 % for determination of absorption coefficient and around 5 % for determination reduced scattering coefficient, as can be seen in Fig. 6b.

Comparison with parameters values extracted from measurements using ISS Imagent frequency domain oxymeter, and their related percent of error, is depicted in Fig. 7a. Meanwhile comparison of derived hemodynamics parameters, i.e. the tissue oxygen saturation (StO_2) and the total hemoglobin concentration (tHB) can be depicted in Fig. 7b.



(a) (b)

Fig. 6. (a) Broadband profile of the extracted absorption coefficient μ_a (λ) and reduced scattering coefficient μ'_S (λ), (b) Precision level of extracted absorption coefficient μ_a (λ) and reduced scattering coefficient μ'_S (λ) (from 5 consecutive reflectance measurements).



(a) (b)

Fig. 7. (a) Broadband profile of the extracted absorption coefficient μ_a (λ) and reduced scattering coefficient μ'_S (λ) for two measurement positions and comparison with ones obtained using Imagent ISS oxymeter, (b) Comparison of derived hemodynamics parameters StO_2 (%) and tHb [µM] with ones measured using Imagent ISS oxymeter.

Extraction of parameters from prepared Intralipid tissue phantom is accomplished using tissue model as follow:

$$\mu_{a,IL}(\lambda) = f_w \, \mu_{water}(\lambda) + f_{absorber} \, \mu_{absorber}(\lambda) \quad (11)$$

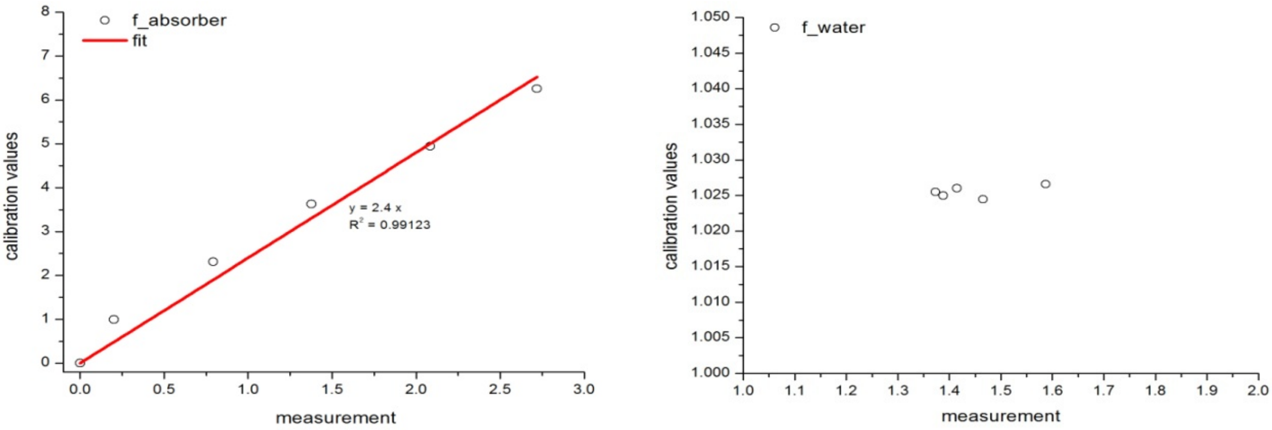
$$\mu'_s(\lambda) = A\lambda^{-B} \quad (12)$$

Parameters extracted from the diffuse reflectance measurements are calibrated with the ones that are fitted from the prescribed values prepared from standard solutions. These values are given in Table 1.

Equating parameters extracted from the diffuse reflectance measurements to the ones fitted from the prescribed values prepared from standard solutions can be seen in following Figs. 8a and 8b.

Table 1. Prescribed values prepared from standard solutions (all values are in cm^{-1} units).

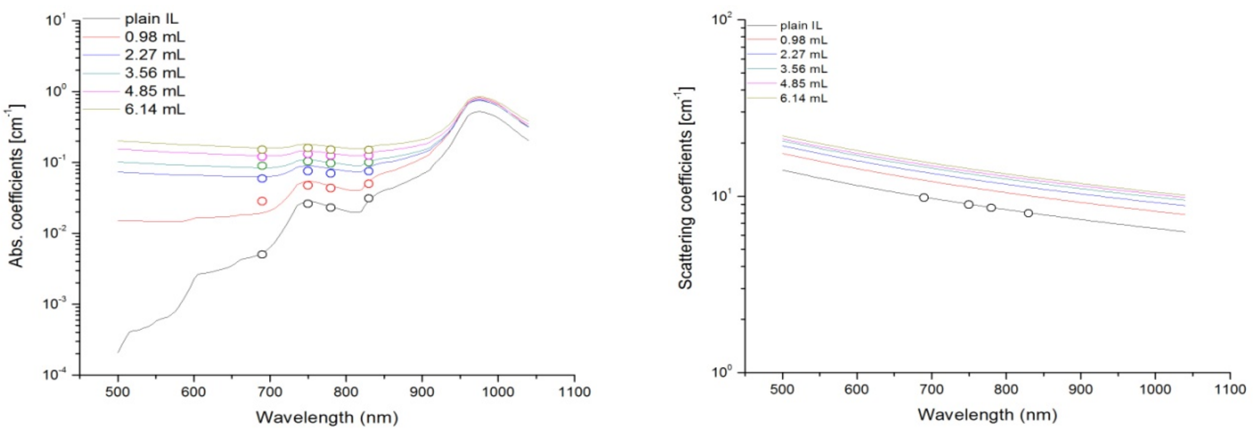
Wavelength (nm)	Plain IL		μ_a for additional drops of India Ink (10 % vol)				
	μ_a	μ_S	0.98 mL	2.27 mL	3.56 mL	4.85 mL	6.14 mL
690 nm	0.005	9.810	0.028	0.059	0.090	0.120	0.151
750 nm	0.026	8.962	0.047	0.075	0.103	0.131	0.159
780 nm	0.023	8.580	0.043	0.070	0.097	0.124	0.151
830 nm	0.031	8.000	0.050	0.075	0.100	0.125	0.150



(a) (b)

Fig. 8. Relationship between parameters extracted from diffuse reflectance measurements to the ones fitted from the prescribed values prepared from standard solutions (a) absorber component (India ink) and (b) water component.

The parameter from absorber component (ink) shows a good linear correlation with a slope of 2.4, meanwhile the parameter from water component (and also the scattering parameters: *A* and *B*) show no correlation at all. Using these results, the calibrated broadband values for both μ_a (λ) and μ'_S (λ) can be plotted in Fig. 9a and 9b.



(a) (b)

Fig. 9. (a) Lines: values of calibrated broadband absorption coefficients μ_a (λ), open circles: prescribed values prepared from standard solutions. (b) lines: values of calibrated broadband reduced scattering coefficients μ'_S (λ), open circle: prescribed value prepared from standard solutions.

Fig. 9 (b) shows that the subsequent additional drops of India ink contribute to the increasing values of reduced scattering coefficients $\mu'_S(\lambda)$, as indicated by Di Ninni [21].

Comparison of accuracy level of broadband $\mu_a(\lambda)$ extraction, calculated relative to the prescribed values of prepared liquid tissue phantom, before and after including the calibration coefficient can be given in Table 2 below. The results show that accuracy level of the calibrated broadband extracted $\mu_a(\lambda)$ values are much improved.

Table 2. Comparison of accuracy level from extracted absorption coefficients $\mu_a(\lambda)$ between *before* (**B**) and *after* (**A**) calibration (in terms of percent errors).

Wavelength (nm)	**Additional drops of India Ink (10 % vol)**											
	plain IL		**0.98 mL**		**2.27 mL**		**3.56 mL**		**4.85 mL**		**6.14 mL**	
	B	**A**	**B**	**A**	**B**	**A**	**B**	**A**	**B**	**A**	**B**	**A**
690 nm	160	**5.6**	54.6	**31.2**	55.9	**6.5**	56.4	**6.1**	53.7	**2.8**	53.0	**5.8**
750 nm	165.4	**9.7**	46.7	**15.4**	25.7	**19.9**	34.2	**5.3**	36.7	**10.7**	38	**13.0**
780 nm	152.2	**3**	32.9	**7.6**	30.2	**16.0**	38.1	**2.4**	39.7	**8.5**	40.6	**11.0**
830 nm	125.8	**6.7**	40.8	**8.2**	26.6	**13.7**	45	**1.7**	37	**7.4**	38	**10.3**

3. Conclusion

A simple and novel scheme for extracting broadband tissue optical properties from broadband diffuse reflectance measurements has been successfully developed. This extraction scheme shows fitting stability, which can be seen from the good repeatability of the parameters extracted. The extracted broadband absorption coefficient values show improving accuracy for spectral region near the isobestic point. Meanwhile the derived hemodynamic parameters of StO_2 and tHb show good conformity with ones derived from measurements using Imagent ISS frequency-domain system. Possible discrepancies might be contributed due to variability in contact between measuring probe and tissue. Calibration using Intralipid tissue simulating phantom shows better improvement in accuracy from extracted absorption coefficients $\mu_a(\lambda)$, as indicated by decreasing percent of error. Further detailed calibration is still required to enhance the accuracy level of the developed extraction scheme, and it will be our focus in the near future. Parallel effort to simulate effect of probe-tissue contact to the extraction accuracy is still being investigated by using Monte Carlo simulation.

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Aims and Scope

Sensors & Transducers Journal (ISSN 1726-5479) provides an advanced forum for the science and technology of physical, chemical sensors and biosensors. It publishes state-of-the-art reviews, regular research and application specific papers, short notes, letters to Editor and sensors related books reviews as well as academic, practical and commercial information of interest to its readership. Because of it is a peer reviewed international journal, papers rapidly published in *Sensors & Transducers Journal* will receive a very high publicity. The journal is published monthly as twelve issues per year by International Frequency Sensor Association (IFSA). In additional, some special sponsored and conference issues published annually. *Sensors & Transducers Journal* is indexed and abstracted very quickly by Chemical Abstracts, IndexCopernicus Journals Master List, Open J-Gate, Google Scholar, etc. Since 2011 the journal is covered and indexed (including a Scopus, Embase, Engineering Village and Reaxys) in Elsevier products.

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