

MEASUREMENT OF THE OPTICAL PROPERTIES OF THE ADULT HUMAN HEAD WITH SPATIALLY RESOLVED SPECTROSCOPY AND CHANGES OF POSTURE

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1. INTRODUCTION

Absolute optical properties (i.e., absorption and reduced scattering coefficients, μ_a and μ_s) of human tissues such as the head, calf and arm have been measured using near-infrared (NIR) phase¹, time² or spatially³ resolved spectroscopy (SRS) systems. While a simple continuous-wave (CW) system can measure $\Delta\mu_a$, absolute μ_a cannot be easily measured because of the complex geometry in which measurements are made in tissues. It has been shown that the SRS technique can be used to calculate a scaled absolute μ_a , i.e. $\mu_s'\mu_a$ where μ_s' is considered as a time-invariant scaling factor⁴. This paper suggests a way to use a commercially available spectrometer, namely the NIRO-300 (Hamamatsu KK.) which has both CW and SRS capabilities, to calibrate an absolute μ_a based on the changes of μ_a (i.e., $\Delta\mu_a$, calculated from the CW data and a modified Beer-Lambert law) and the scaled μ_a (i.e., $\mu_s'\mu_a$ calculated from the SRS data). Using changes of posture from the supine to the head up position, the absolute optical properties of 15 adult human heads were calculated. Issues of errors due to the inhomogeneity in real tissues and methods to minimise them are also discussed.

2. THEORY

Based on the modified Beer-Lambert law (BL), $\Delta\mu_a^{BL}$ can be calculated from a measurement of a change of attenuation (ΔA):

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$$\Delta\mu_a^{BL} = \frac{\Delta A \times \log_e 10}{\rho \times DPF} \quad (\text{mm}^{-1}) \quad (1)$$

where ρ is the optode spacing and the DPF is the differential pathlength factor. When ΔA is measured with the log base of 10, the scaling factor $\log_e 10$ needs to be introduced to convert equation (1) to the log base of e with which μ_a is defined in the diffusion equation.

Based on the semi-infinite half-space geometry, the solution of the diffusion equation in a CW system can be differentiated with respect to the optode spacing, resulting in an expression for μ_a^{SRS} as a linear function of μ_s' as shown in equation (2)⁴.

$$\mu_s' \mu_a^{SRS} = \frac{1}{3} \left(\log_e 10 \frac{\partial A}{\partial \rho} - \frac{2}{\rho} \right)^2 \quad (\text{mm}^{-2}) \quad (2)$$

where $\partial A / \partial \rho$ is the attenuation slope measured with multiple detectors and has a log base of 10. Since both $\Delta\mu_a^{BL}$ in equation (1) and μ_a^{SRS} in equation (2) correspond to essentially the same haemoglobin dependent chromophore, a linear equation (c.f. $y = mx+c$) can be formed considering those two equations:

$$\mu_s' \mu_a^{SRS} = \mu_s' \Delta\mu_a^{BL} + \mu_s' \mu_a^{base} \quad (3)$$

where μ_a^{base} is the baseline value from which subsequent μ_a^{BL} are subtracted to form $\Delta\mu_a^{BL}$. With a range of haemoglobin concentrations, one can plot $\Delta\mu_a^{BL}$ against $\mu_s' \mu_a^{SRS}$ and a straight line can be fitted by linear regression. The slope of the straight line is μ_s' . The μ_a^{SRS} can then be separated into the contributing components:

$$\mu_a^{SRS}(\lambda) = \alpha_{HHb}(\lambda)C_{HHb} + \alpha_{HbO_2}(\lambda)C_{HbO_2} + G \quad (4)$$

where α_{HHb} , α_{HbO_2} , C_{HHb} and C_{HbO_2} are the specific absorption coefficients and concentrations of deoxy- and oxy-haemoglobins, respectively, and G includes all background absorbers and errors due to deviations from a simple homogeneous diffusion model (e.g. real tissue heterogeneity or absorber distribution). In order to minimise the effect of G , equation (3) can be rewritten using the difference of μ_a between two wavelengths, i.e. λ_i and λ_j :

$$\begin{aligned} \mu_s'(\lambda_j)[\mu_a^{SRS}(\lambda_j) - k(\lambda_i, \lambda_j)\mu_a^{SRS}(\lambda_i)] \\ = \mu_s'(\lambda_j)[\Delta\mu_a^{BL}(\lambda_j) - k(\lambda_i, \lambda_j)\Delta\mu_a^{BL}(\lambda_i)] \\ + \mu_s'(\lambda_j)[\mu_a^{base}(\lambda_j) - k(\lambda_i, \lambda_j)\mu_a^{base}(\lambda_i)] \end{aligned} \quad (5)$$

where $k(\lambda_i, \lambda_j)$ is a scaling factor correcting for the wavelength dependence of μ_s' and is

$$\text{defined as :} \quad k(\lambda_i, \lambda_j) = \frac{\hat{\mu}_s'(\lambda_i)}{\hat{\mu}_s'(\lambda_j)} \quad (6)$$

and

$$\hat{\mu}_s'(\lambda) = a\lambda + b \quad (7)$$

where $\hat{\mu}_s'(\lambda)$ has previously been estimated experimentally² and $a = -6.5 \times 10^{-4} \text{ mm}^{-1} \text{ nm}^{-1}$ and $b = 1.45 \text{ mm}^{-1}$. Re-writing equation (5) using simplified symbols:

$$\mu_s'(\lambda_j)\mu_a^{SRS}(\Delta\lambda_{ji}) = \mu_s'(\lambda_j)\Delta\mu_a^{BL}(\Delta\lambda_{ji}) + \mu_s'(\lambda_j)\mu_a^{base}(\Delta\lambda_{ji}) \quad (8)$$

In summary, when both $\Delta\mu_a^{BL}$ and μ_a^{SRS} are collected using a spectrometer with three wavelengths, the following estimation procedures for μ_s' and μ_a can be carried out:

1. Estimation of $\mu_s'(\lambda_j)$ by a linear regression between $\Delta\mu_a^{BL}(\Delta\lambda_{ji})$ and $\mu_s'(\lambda_j)\mu_a^{SRS}(\Delta\lambda_{ji})$:

$$\mu_s'(\lambda_j) = \frac{\mu_s'(\lambda_j) \Delta \mu_a^{SRS}(\Delta \lambda_{ji})}{\Delta \mu_a^{BL}(\Delta \lambda_{ji})} \quad (9)$$

2. Repeat the procedure for all three λ to select the $\mu_s'(\lambda)$ giving the highest correlation coefficient between $\Delta \mu_a^{BL}(\Delta \lambda_{ji})$ and $\mu_s' \mu_a^{SRS}(\Delta \lambda_{ji})$, i.e. $\mu_s'(\lambda^*)$.

3. Scaling of the remaining two μ_s' at the other two wavelengths according to :

$$\mu_s'(\lambda) = \frac{a\lambda + b}{a\lambda^* + b} \mu_s'(\lambda^*) \quad (10)$$

4. Estimation of the μ_a at the three wavelengths by simple substitutions :

$$\mu_a(\lambda) = \frac{\mu_s'(\lambda) \mu_a^{SRS}(\lambda)}{\mu_s'(\lambda)} \quad (11)$$

5. Since the absolute μ_a at three wavelengths are found, conversion to absolute haemoglobin concentration can also be carried out. Using the model given in (4), one can write the following expression to minimise G :

$$\mu_a(\lambda_i) - \mu_a(\lambda_j) = [\alpha_{HHb}(\lambda_i) - \alpha_{HHb}(\lambda_j)] C_{HHb} + [\alpha_{HbO_2}(\lambda_i) - \alpha_{HbO_2}(\lambda_j)] C_{HbO_2} \quad (12)$$

With three wavelengths available, two independent equations with the same form as equation (12) can be written, enabling the calculation of C_{HHb} , C_{HbO_2} and the total haemoglobin concentration, C_{HbT} ($= C_{HHb} + C_{HbO_2}$). When C_{Hbt} is available, the cerebral blood volume (CBV) can also be calculated with the following formula⁶:

$$CBV = \frac{C_{HbT} \times MW_{Hb} \times 10^{-4}}{d_t \times Hb_t \times CLVHR} \quad (\text{ml}/100\text{g}) \quad (13)$$

where $MW_{Hb} = 64500$ g is the molecular weight of haemoglobin, $d_t = 1.05$ g/ml is the cerebral tissue density, Hb_t (g/dl) is the haemoglobin concentration obtained from a venous sample, and $CLVHR = 0.69$ is the cerebral large-to-small vessel haematocrit ratio.

3. METHODS

3.1 Subjects and experiment protocols

Fifteen subjects with primary autonomic failure were involved in this study. The patients were aged between 42 and 78 with a median age of 62. The study was approved by the hospital ethics committee. This group of patients was chosen because their cerebral blood volumes were expected to change significantly upon a change of posture from the supine to head up position. This was expected to provide a wide range of change in μ_a and hence a better estimate of μ_s' and μ_a using the technique proposed in section 2. An NIR spectrometer with SRS capability (NIRO-300, Hamamatsu Photonics KK) was used in this study. The optical probe was attached to the foreheads of the subjects with an optode spacing of 50 mm. Subjects lay on a tilt table in the supine position at the beginning of the experiment. They were then tilted head up passively to 60° for 5 to 10 minutes, depending on how long they remained asymptomatic, before being lowered back down to the horizontal position.

3.2 Data analysis

A five minute section of data was analysed during the head up tilt using the method discussed in section 2. The value of DPF was chosen according to the age-dependent formula given in a previous study⁵:

$$DPF = 5.13 + 0.07 \times AGE^{0.81} \quad (13)$$

4. RESULTS

An example of the $\Delta\mu_a^{BL}(\Delta\lambda_{21})$ and $\mu_s'(\lambda_2)\mu_a^{SRS}(\Delta\lambda_{21})$ signals from one subject is shown in Figure.1 where $\lambda_1=775\text{nm}$ and $\lambda_2=813\text{nm}$. It can be seen that both signals dropped gradually during the head up tilt and recovered after the supine position was resumed. In Figure.2, the correlation between the $\Delta\mu_a^{BL}(\Delta\lambda_{21})$ and $\mu_s'(\lambda_2)\mu_a^{SRS}(\Delta\lambda_{21})$ data can be seen. The slope of the regressed straight line is $\mu_s'(\lambda_2)$. The mean and standard deviation of μ_s' and μ_a at three wavelengths estimated from 15 subjects are shown in Table 1. The mean and standard deviation of the estimated C_{HbT} are $43.00 \pm 14.74 \mu\text{M}$.

Table 1. The mean and standard deviation of μ_s' and μ_a at three wavelengths (n=15)

	$\mu_s'(775\text{nm})$ (mm-1)	$\mu_s'(813\text{nm})$ (mm-1)	$\mu_s'(853\text{nm})$ (mm-1)	$\mu_a(775\text{nm})$ (mm-1)	$\mu_a(813\text{nm})$ (mm-1)	$\mu_a(853\text{nm})$ (mm-1)
mean	1.047	1.020	0.993	0.008	0.009	0.010
std.	± 0.575	± 0.560	± 0.545	± 0.003	± 0.003	± 0.004

5. DISCUSSION AND CONCLUSIONS

The estimated mean μ_s' is comparable to the previously published *in vivo* result using a time-resolved system² ($\mu_s'(800\text{ nm}) = 0.94 \pm 0.07 \text{ mm}^{-1}$), while the estimated mean μ_a is lower than that obtained in the same study² ($\mu_a(800\text{ nm}) = 0.016 \pm 0.001 \text{ mm}^{-1}$). The estimation technique proposed here relies on the accurate measurement of the scaled μ_a and change of μ_a which in turn are based on the SRS and modified Beer-Lambert law respectively. Difficulties with this technique include limitations in the assumption of homogeneity used in the SRS calculation, and inter-subject variability in DPF . The modified Beer-Lambert law requires the DPF , which is currently pre-set according to previously published results, but in reality likely to be subject dependent. The DPF acts like a scaling factor which directly affects the estimated μ_s' and μ_a . These limitations partly explain the large standard deviation of μ_s' and the discrepancies between our μ_a result and the previously published result. The estimated mean CBV is comparable with the result previously published using a CW system⁶ ($2.85 \pm 0.97 \text{ ml}/100\text{g}$), but is smaller than that obtained from a SPECT study⁷ ($4.81 \pm 0.37 \text{ ml}/100\text{g}$) for regional brain tissues. The SPECT result focused on the region of the actual brain whereas near-infrared systems probe the whole head including scalp, skull, and brain, resulting in an averaged blood volume, which may be lower than the brain blood volume alone. In comparison with the oxygen swing NIR technique proposed before⁶, this method described in this

paper requires only a simple posture change for the estimation of CBV , which greatly simplifies the measurement procedure.

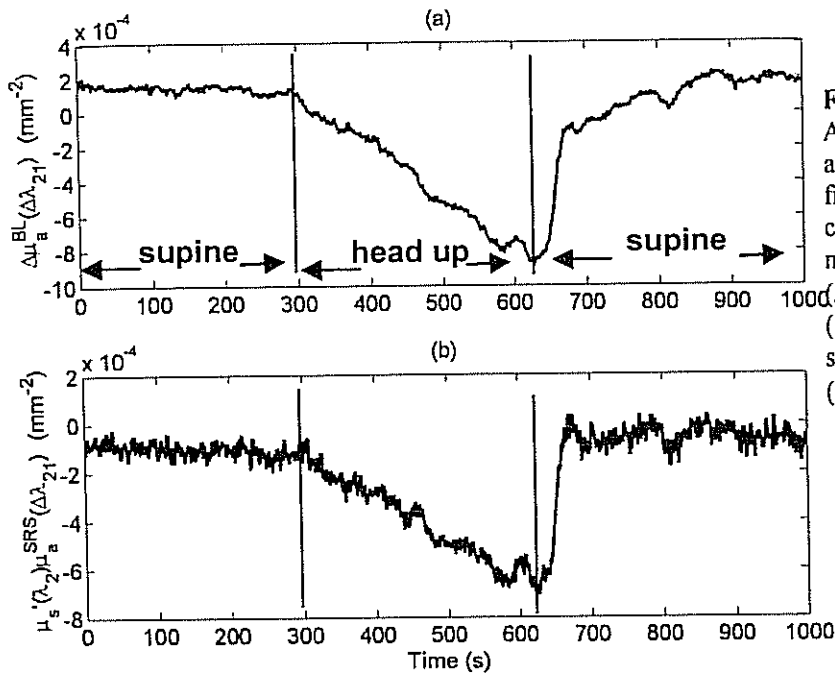


Figure 1
An example of the absorption coefficients from one subject calculated using (a) the modified Beer Lambert law ($\Delta\mu_a^{BL}(\Delta\lambda_{21})$) and (b) the spatially resolved spectroscopy ($\mu_s'(\lambda_2)\mu_a^{SRS}(\Delta\lambda_{21})$)

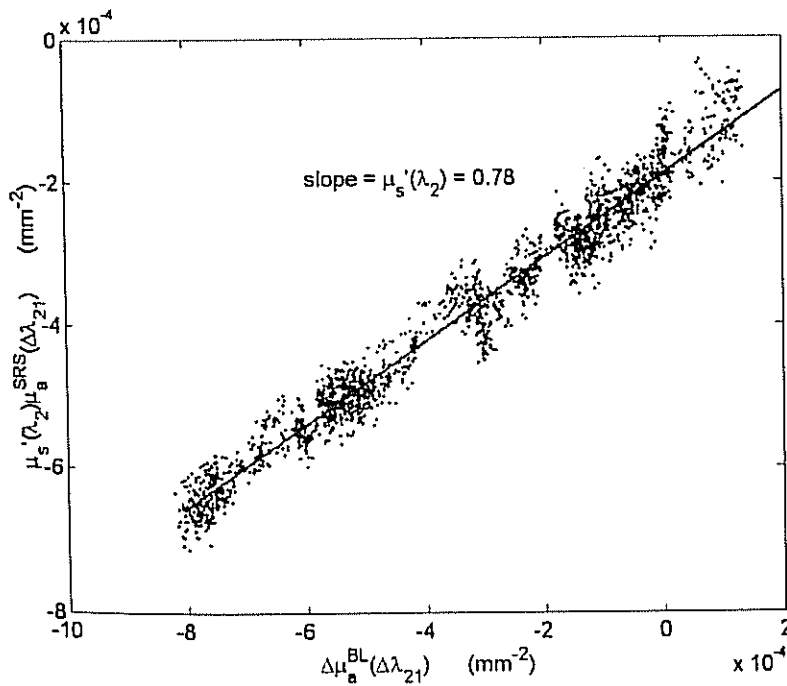


Figure 2
Correlation between the absorption coefficients calculated using the modified Beer Lambert law ($\Delta\mu_a^{BL}(\Delta\lambda_{21})$) and using the spatially resolved spectroscopy ($\mu_s'(\lambda_2)\mu_a^{SRS}(\Delta\lambda_{21})$)

6. ACKNOWLEDGEMENTS

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7. REFERENCES

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