

## THE EFFECT OF OPTODE POSITIONING ON OPTICAL PATHLENGTH IN NEAR INFRARED SPECTROSCOPY OF BRAIN.

P. van der Zee, S.R. Arridge, M. Cope, D.T. Delpy.

Department of Medical Physics & Bioengineering  
University College London  
1st Floor Shropshire House  
11-20 Capper Street  
London WC1E 6JA

### INTRODUCTION

The use of optical spectroscopy for the non-invasive monitoring of tissue oxygenation and metabolism is well established (Chance et al., 1975). Historically because of the high absorption by tissue of light in the visible range, optical monitoring was often restricted to measurements of reflected light (Jöbsis et al., 1977). Subsequently Jöbsis showed that by using near infrared light (NIR), tissue absorption became sufficiently low to make transillumination of the cat head possible (Jöbsis, 1977). In the near infrared region (700-1300 nm) there is sufficient spectral information available to permit changes in the concentration of haemoglobin and cytochrome  $aa_3$  to be calculated, and hence changes in the oxygenation state of the brain (Brazy et al., 1985, 1986; Ferrari et al., 1986; Fox et al., 1985). This technique is now used routinely to monitor cerebral oxygenation and haemodynamics in the human newborn infant (Wyatt et al., 1986; Edwards et al., 1988), using an instrument designed to transilluminate the heads of most newborn infants (Cope and Delpy, 1988). This instrument allows for measurements through heads up to 8-9 cm in diameter.

In order to quantitate NIRS haemoglobin and cytochrome  $aa_3$  data, it is normally necessary to know the optical pathlength of the light in the tissues. Due to the high amount of light scattering in tissues this is appreciably larger than the interoptode spacing. For the case of transillumination through a slab of tissue we have shown (Delpy et al., 1988) that the effective optical pathlength through the tissue is equal to the interoptode spacing multiplied by a near constant factor (the differential pathlength factor or DPF) which is dependent upon the optical characteristics of the tissue. The DPF, which relates a change in measured light transmission to a change in true absorption, can be derived from a measurement of the time of flight for ultrashort optical pulses through the tissue. For the above slab geometry, using both a Monte Carlo model and measurements on a phantom, we were able to show that the DPF corresponded to the mean transit time of the photons emerging from the tissue. By applying this time of flight technique to the case of the transilluminated heads of pre term infants, the DPF has been found to be 4.4 (Wyatt et al., 1989). In a further study in animals, the change of the DPF with absorption in the brain was investigated (Delpy et al., 1989). The DPF was found to change by 30%/OD/cm, the change being approximately linear over the whole physiological range from  $FiO_2 = 100\%$  to death. For the normal physiological range of oxygenation and blood volume, the maximum

absorption change observed in the brain, both in animals and in human studies, is typically 0.2 OD/cm, resulting in a DPF change of less than 10%.

Although the DPF, as determined by the time of flight (TOF) measurements, has already been applied to data from rat and human infant brain, the validity of the mean time of flight as a measure of pathlength has only been verified so far for a slab geometry. Also it is not always possible to transilluminate the head of larger infants, and in these cases the optodes must be positioned such that they only partially illuminate the tissues. In the case of large full term infants it is often necessary to position one optode over the fontanelle with the other on the side of the head. In such cases, the angle between the optodes may be  $90^\circ$  or less. The questions that must be asked in these circumstances are firstly whether the DPF as determined from optical time of flight measurements applies to a spherical geometry, which more closely resembles the shape of the head, and secondly whether the calculated DPF is valid for arbitrary optode positioning on the (approximately) spherical head.

### MODELLING OF LIGHT TRANSPORT IN TISSUE.

To investigate the effects of optode geometry on the effective pathlength for light going through highly scattering tissues, a Monte Carlo model for light transport in tissue was used (van der Zee and Delpy, 1987). The model gives a full three dimensional simulation of light transport. The scattering phase function of the tissue used in the model is based on an experimentally measured phase function for rat brain at 783 nm (van der Zee and Delpy, 1988). Specular reflection and refraction at the tissue boundary, due to differences in refractive index were taken into account.

In this simulation, a perfectly collimated, infinitesimally narrow and infinitely short pulse of light was incident at  $(0,0,-r)$ , on a sphere of radius  $r$  (Figure 1). The model keeps track of the photons as they pass through the sphere. The angular position on the sphere,  $\Theta$ , and the transit time  $t$  are recorded for each photon that emerges. The data is collected into ten annular rings for values of  $\Theta$  in  $20^\circ$  intervals centred on:  $0^\circ$ ,  $20^\circ$ ,  $40^\circ$ ,  $60^\circ$ ,  $80^\circ$ ,  $100^\circ$ ,  $120^\circ$ ,  $140^\circ$ ,  $160^\circ$  and  $180^\circ$ . Within each of these annular rings a distribution of transit times is obtained, and the mean of this time ( $t_{\text{mean}}$ ) is determined.

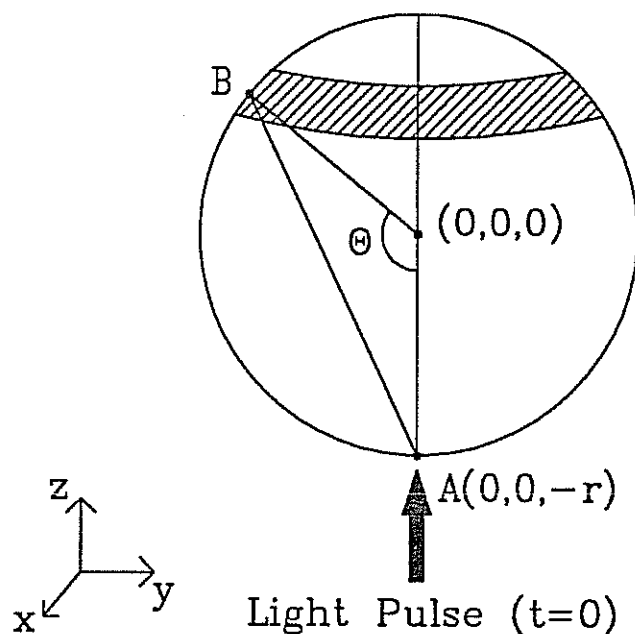


Figure 1. Geometry for the model. Light enters at the bottom of the sphere. The exit position is specified by the angle  $\Theta$ . The optode spacing equals AB.

Also the intensity of the exiting light in each ring, normalised for the surface area of that ring is calculated. For the interoptode spacing, the cord length AB from input to output site is used ( $AB = 2r \cdot \sin(\Theta/2)$ ). The DPF for a particular angle is obtained as follows :

$$DPF(\Theta) = (t_{\text{mean}}(\Theta) \cdot c/n) / AB.$$

Here  $c$  is the velocity of light in vacuo and  $n$  is the tissue refractive index, taken to be 1.4 (Gahm et al., 1986). For each ring the change in attenuation per unit change in absorption, normalised for the chordlength AB, is the incremental Beer Lambert pathlength factor for that angle (ie  $(\delta OD(\Theta)/\delta \mu_a)/AB$ ). The parameters for which the model was run were: sphere radius  $r=15\text{mm}$ , scattering coefficient  $\mu_s=10\text{mm}^{-1}$  (1/e) and for a range of absorption coefficients :  $\mu_a=0.01, 0.02, 0.04, 0.06, 0.08, 0.10\text{mm}^{-1}$  (1/e). The total number of detected photons for each absorption coefficient was 1.2 million.

## RESULTS

Figures 2a, b, c show the calculated differential pathlength factor derived from the mean of the time of flight plotted against the incremental Beer Lambert pathlength factor determined from  $(\delta OD/\delta \mu_a)/AB$ , for exit angles of respectively 130-180°, 70-130° and 10-70°. The line of identity is also shown.

In Figures 3a and 3b, the DPF is plotted as a function of  $\Theta$  for absorption coefficients  $\mu_a$  of 0.01 to 0.10  $\text{mm}^{-1}$ . The DPF can be seen to decrease with increasing absorption, as reported previously (Delpy et al., 1989), furthermore there is a variation with  $\Theta$ .

In Figure 4, the DPF, normalised to unity at  $\mu_a = 0.010$  is plotted as function of absorption coefficient,  $\mu_a$ , for different values of  $\Theta$ .

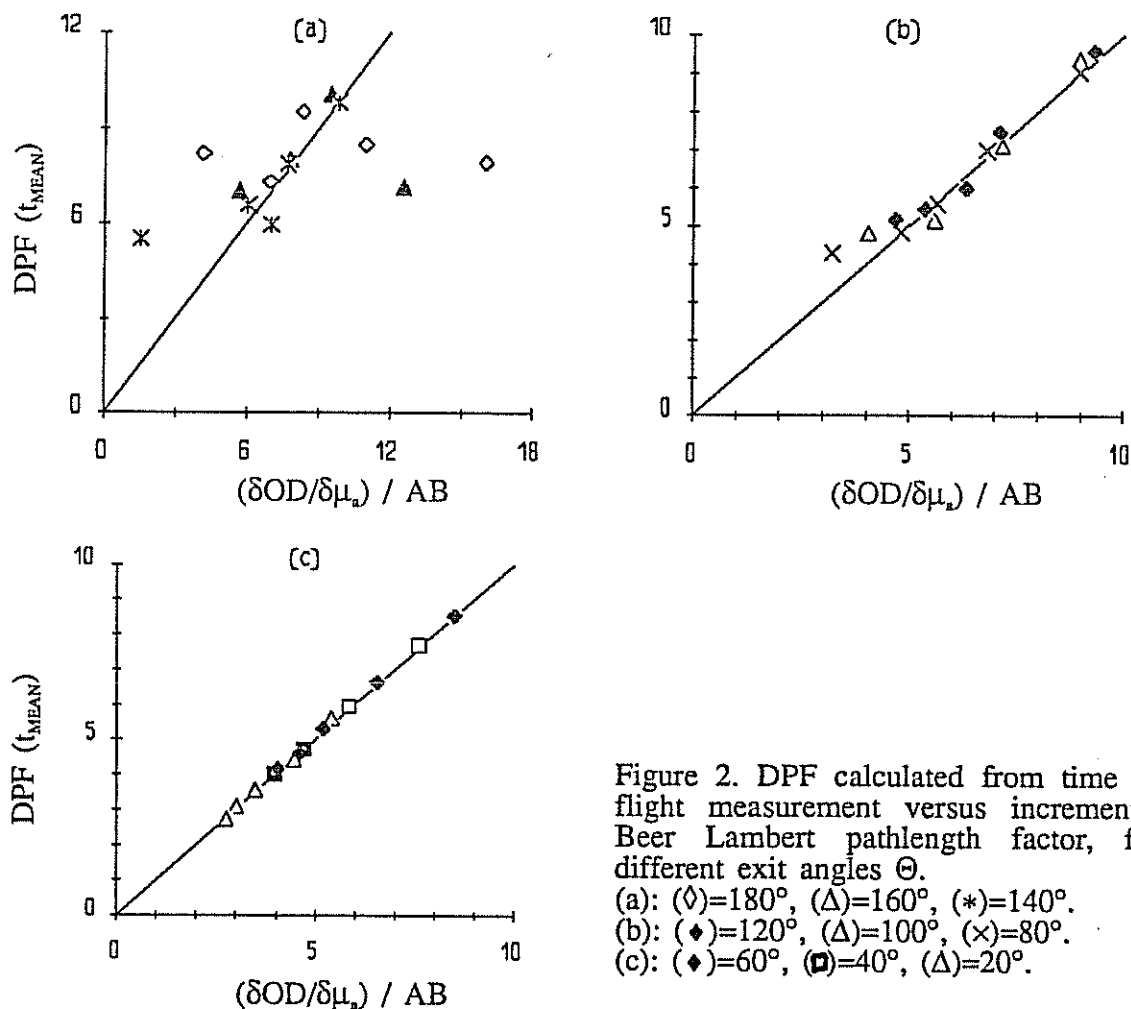


Figure 2. DPF calculated from time of flight measurement versus incremental Beer Lambert pathlength factor, for different exit angles  $\Theta$ .

(a): ( $\diamond$ )=180°, ( $\Delta$ )=160°, (\*)=140°.

(b): ( $\diamond$ )=120°, ( $\Delta$ )=100°, ( $\times$ )=80°.

(c): ( $\diamond$ )=60°, ( $\square$ )=40°, ( $\Delta$ )=20°.

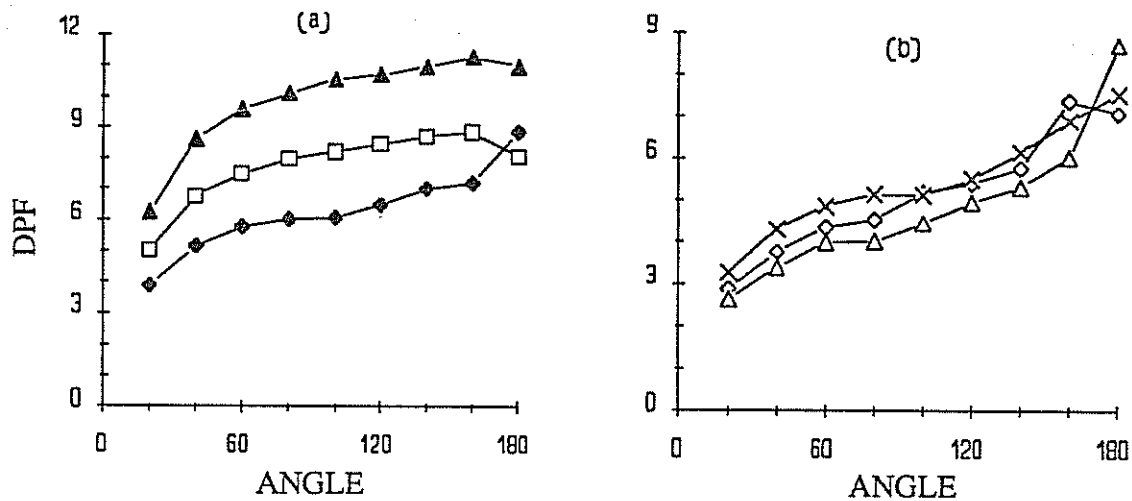


Figure 3. Differential pathlength factor (DPF) as a function of angle for different absorption coefficients,  $\mu_a$ . (a): ( $\Delta$ )=0.01, ( $\square$ )=0.02, ( $\diamond$ )=0.04. (b): ( $\times$ )=0.06, ( $\circ$ )=0.08, ( $\Delta$ )=0.10  $\text{mm}^{-1}$ .

DISCUSSION

The first question asked concerned the validity for a spherical geometry of the use of the mean of the optical time of flight to determine the differential pathlength factor (DPF) relating variations in attenuation to true absorption changes. The data in Figures 2a,b & c show a good linear relationship with unity slope and zero intercept between the DPF derived from the mean transit time and the incremental Beer Lambert pathlength factor,  $(\delta\text{OD}/\delta\mu_a)/\text{AB}$ . This is seen clearly for the smaller exit angles. For the larger angles there is a larger spread in the data points due to lack in statistics, but the one to one relationship between DPF and  $(\delta\text{OD}/\delta\mu_a)/\text{AB}$  still holds true. The change in the DPF with  $\Theta$ , as seen in Figures 3a and 3b, shows a decrease in DPF in going from  $180^\circ$ , full transillumination, to  $20^\circ$ , close to pure reflection. Between  $180^\circ$  and  $60^\circ$  this change is quite gradual, changing by approximately 2 DPF over the range. It is interesting to note that the DPF change is approximately independent of absorption and

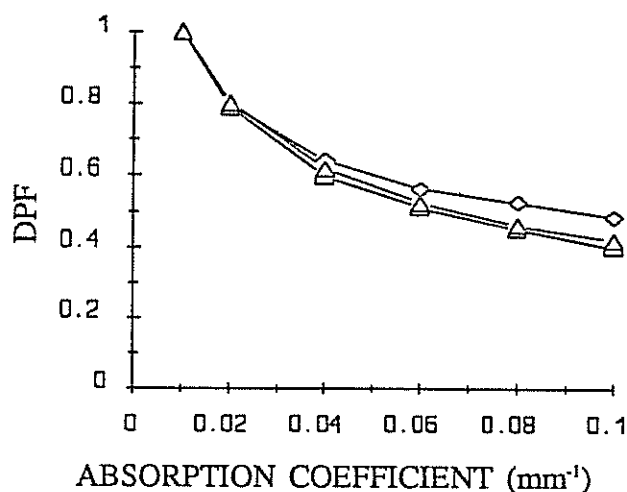


Figure 4. DPF as a function of absorption coefficient for different angles,  $\Theta$ . The data are normalised to the DPF's at  $\mu_a=0.01 \text{ mm}^{-1}$ . ( $\diamond$ ):  $\Theta=140^\circ$ , ( $\square$ ):  $\Theta=80^\circ$ , ( $\Delta$ ):  $\Theta=20^\circ$ .

thus the percentage change will be greater for the higher absorption values. The typical value for blood perfused brain would be 2.7%/10°. At  $\Theta$  less than 60° there is a more rapid decrease in DPF with angle, particularly for the case of low absorption. This could lead to large errors in the calculation of chromophore concentration from measurements made in reflection mode. In particular, reflection measurements made on tissues of low absorption could be prone to error due to the optical pathlength changing substantially with variations in tissue absorption (eg fluorocarbon perfused rat brain, where the total absorption coefficient is only 0.018mm<sup>-1</sup> at 780nm, for 20µMolar of oxygenated cytochrome aa<sub>3</sub> and 80% water content). A possible explanation for the lower DPF values at decreasing angle is that photons are lost from the sphere. With decreasing angle  $\Theta$  (Figure 1), photons are unable to take long paths on the outside of the sphere and are therefore lost to free space. This leads to a shortening of the mean time relative to the chord length AB. This argument is consistent with the rapid decrease in DPF at very small angles.

The relative change in the DPF with absorption, as shown in figure 4, is seen to be largest at low absorption, and for smaller angles, with remarkably little difference between the results for  $\Theta=80^\circ$  and  $\Theta=20^\circ$ .

## CONCLUSIONS

Using a computer simulation for light transport in tissue it has been shown that the mean of the time of flight for a short pulse of light through a sphere of tissue is an accurate measure of the DPF, the factor used in converting changes in attenuation to true concentration changes. This DPF changes with angular position on the sphere. For the range of absorption coefficients found in normally perfused brain, and for optode geometries varying from 180° to 60°, use of the same DPF as a multiplier of the direct interoptode spacing leads to an error in the calculated concentration changes of up to 35%. Therefore for more precise calculations, the DPF should be varied according to the optode geometry employed in the measurements. For angles of less than 60° there is a more rapid change in DPF with both angle and tissue absorption coefficient precluding the use of a simple constant pathlength factor. This leads to a difficulty in the interpretation of reflection type measurements in the near infrared region. These problems could of course be overcome by incorporating a time of flight measurement facility into the tissue spectrometer.

## ACKNOWLEDGEMENTS

This work was carried out with funding provided by the SERC, the Wellcome Trust and Hamamatsu Photonics K.K.

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## KEYWORDS

Infrared Spectroscopy, optical pathlength, Monte Carlo, Differential Pathlength Factor, modelling, sphere, tissue spectroscopy.