Determination of optical properties of tissue and other bio-materials

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INTRODUCTION

A major step in improving medicine is identifying different, effective and efficient methods for diagnosis and treatment which also promote patient comfort. This has led to current trends in healthcare such as non-invasive medical techniques and devices that often utilise some form of light, a laser, LED etc. Therefore knowledge about the interaction between the tissue and these light sources is important. Spiritually many cultures and religions refer to the curative effects of light and scientifically studies have substantiated that light of specific wavelengths can cause healing effects.[1,2,3]

Areas where these non-invasive modalities are applied include photodynamic therapy (PDT) and wound healing. For optimised effects in tissue optical studies, it is important that in the process of treatment, healthy tissue sustains minimal damage[4]. Thus, it is important to calculate and model factors such as the dosage requirements for treatment. Such calculations require the optical properties specifically absorption coefficient (μ_a) and scattering coefficient (μ_s) to be known. The absorption coefficient provides information on the concentration of various chromophores while the scattering coefficient provides information on parameters such as the form, size and concentration of the scattering components. Although some of these properties exist in literature, the composition of tissue varies and, where possible, it is preferred that these properties are measured, in situ. The measurement of optical properties on different South African tissue/skin types can help to optimise techniques or instrumentation specifically for South Africa's diverse population. Some models for such work have recently been studied[5], proving the need to correctly measure and investigate these properties.

The integrating sphere method used in this study requires *in vitro* samples unlike other instrumentation that can do both *in vivo* and *in vitro* samples. It however offers the advantage of providing an accurate lab instrument that is simple to operate and does not require expensive accessories for analysis; a significant factor when hoping to reach most South Africans.

THEORY

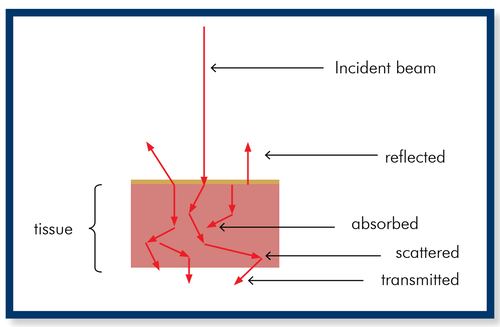

Figure 1: Illustration of the basic phenomena (and model used) accompanying the light-tissue interaction

When light is incident on tissue (as in **Figure 1**), it interacts with the tissue and may be absorbed, scattered through the tissue, be reflected, or be transmitted straight through depending on the tissue's optical properties. For current investigations, μ_a and μ_s' are determined from the measurements using a calibration model and are extracted using the multiple polynomial regression method[6].

MATERIALS AND METHOD

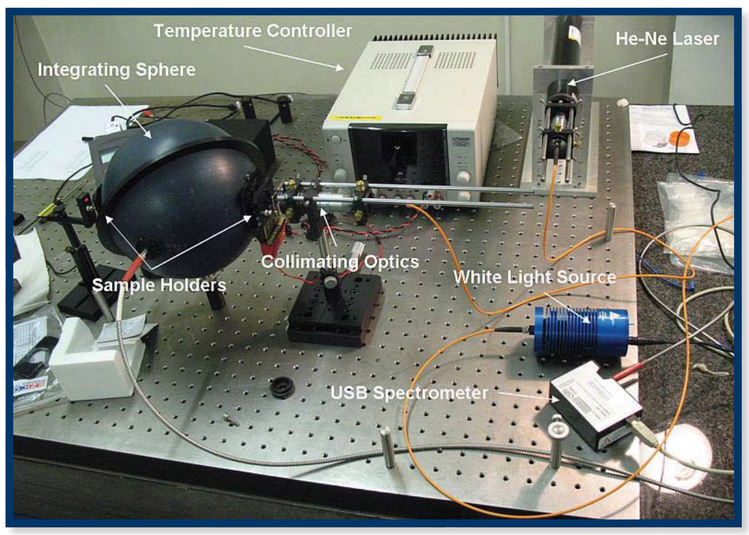

Figure 2: The integrating sphere setup

The integrating sphere comprised a 632.8 nm 7.4 mW He-Ne laser fibre coupled and collimated to a beam diameter of 2-3 mm. A heating system was connected to the sample holders (diameter 25 mm) on the entrance and exit ports of the sphere. A wavelength of 632.8 nm was used, which is a common wavelength used in tissue propagation studies. It falls within the diagnostic and therapeutic window for tissue that lies in the region of 600-1200 nm - where the absorption and scattering properties of tissue dominate over water.

A tissue simulating calibration model was created by constructing different solutions consisting of intralipid and black ink aqueous solutions

RESULTS AND DISCUSSION

Using the prediction algorithm, the inner matrix of the calibration model could be predicted. **Figure 3** illustrates that for this particular calibration model, the higher the order of the polynomial the more accurate the prediction. Noteworthy is that the final calibration model is irregular but maintains low errors of prediction; provided the measured data falls within the range of the model. This is a good indication of the versatility of the model.

Porcine skin was measured directly (dry) and then allowed to hydrate in water for at least 4 hours and measured again. The μ_s' of the 'wet' skin was found to be 25% of the literature value[7] illustrated in **Table 1**. The υ_a for the measured sample is also about half that of the literature values. These results were consistent with previous measurements on another sample of porcine skin measured on the system. The results obtained may be due to different species of pig studied in literature and measured as well as the tissue state (time after excision etc). Also, the properties obtained could represent that of the stratum corneum and not the epidermis as is assumed. The properties of the dry skin could not be extracted with sufficient certainty.

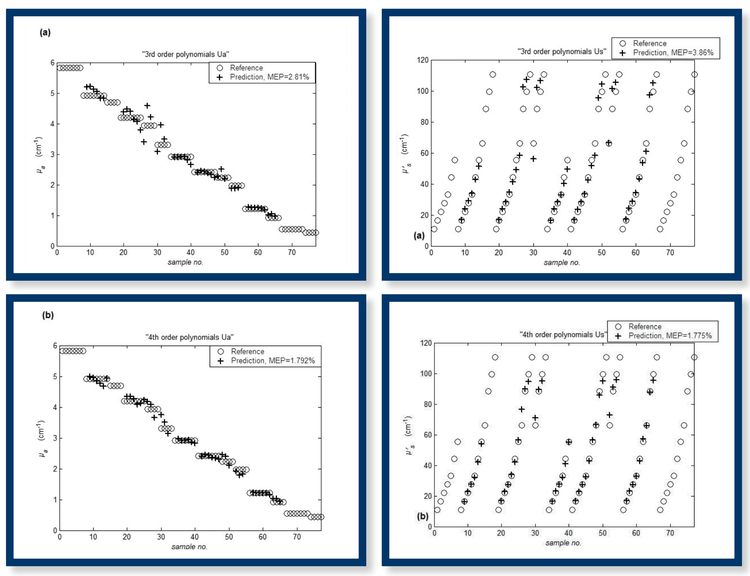

Figure 3: The 3rd and 4th order polynomial predictions of μ_a and μ_s for the calibration model applied (MEP is the mean error prediction)

Table 1: Comparison between literature and measured properties

	υ_a(cm^{-1})	μ_s'(cm^{-1})
Epidermis [7]	1.0±0.1	22.7±0.8
Dermis [7]	0.89±0.1	21.2±0.4
Wet skin (LB)	0.44±0.06	6.67±0.6
Wet fat (underside of skin) (LB)	0.52±0.06	6.36±0.7

The results illustrated the need to measure samples *in situ*. Once determined the optical properties were input to the ASAP (Breault Research Organisation, Inc.) software[5] to model the propagation of light through the different samples.

MODELLING

For the literature values of both porcine and human skin, a 2 layer model with the epidermis and dermis was used. The epidermis was taken to be 0.15mm and the dermis 1.8mm. For the skin + fat model, the skin part was taken to be layer1 2mm and the fat layer2 0.26mm. The quantity measured was for the bulk skin sample and was not separated into the dermis, epidermis and stratum corneum. The absorption was evaluated at different depths into the skin. The human skin using literature[8] values was modeled because pig skin is often taken to be biologically similar to humans. However, the optical properties may differ and this should be noted when transferring tissue optics studies of porcine samples to human models.

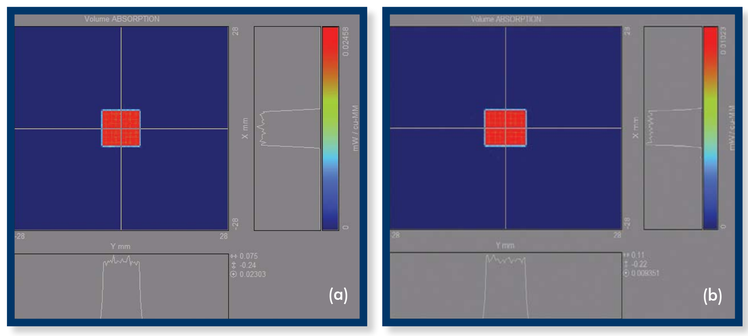

Figure 5: Absorption through layer of skin (a) literature[7] and (b) measured (LB) respectively at a depth of 0.065mm

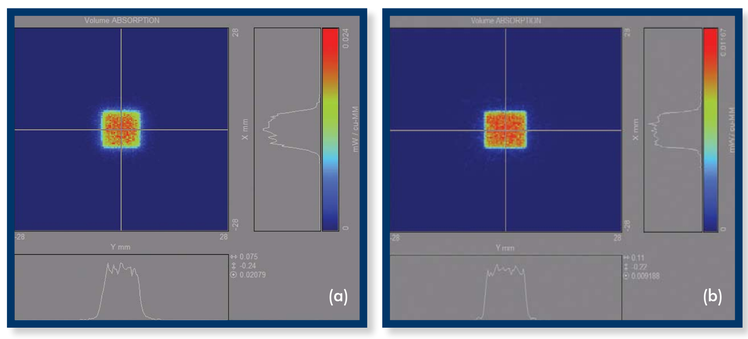

Figure 6: Absorption through skin (a) literature[7] and (b) measured (LB) respectively at a depth of 1.7mm

Figures 5 and **6** illustrate that at the first layer, the distribution of the absorption through the layer is almost the same; the exact absorption differs. However, as the light propagates through the sample, there is more attenuation through the layers of the higher scattering and absorption sample than for the measured sample. In the second instance, the light appears less diffusively scattered.

Extending the frontier into the world of non-invasive medical techniques using lasers on tissue and skin

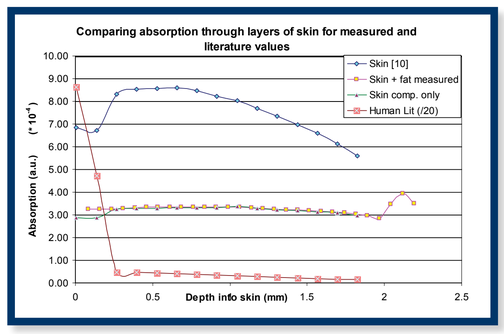

Figure 7: Absorption at different depths of skin for pig and human

Figure 7, obtained from data extracted from the models reinforces these findings. For the model which included the 'fat', the model defines the point at which the fat is present revealing a distinct change in the absorption even though the change in the absorption property between the epidermis and fat is ~ 0.1 cm^{-1}. These changes are significant in detecting an abnormality in tissue. The results also indicate that within the first few mm in the pig tissue the absorption is at a minimum, then increases at the deeper levels of tissue and again decreases towards the latter part of the tissue. Paper[5] further elaborates on how these properties specifically impact the South African diaspora and its different skin types.

CONCLUSIONS

The current setup for the measurement of tissue optical properties using the Integrating sphere has been successfully calibrated and used to determine the optical properties of some tissue and other samples.

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