Light-tissue interaction modelling using COMSOL Multiphysics for multi-layered skin tissues

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Abstract: Diagnosis of soft tissue abnormalities normally follows invasive tissue biopsy or blood analysis. Diffuse Optical Spectroscopy (DOS) is a non-invasive alternative to tissue biopsy that generally employs the reflected or transmitted light between multiple source-detector pairs on the tissue surface to reconstruct the distributions of the optical properties. The variations of these properties in turn is related to the physiological or pathological status of the target tissue. In this work, we modelled a multi-layered skin tissue in COMSOL and studied the effect of light-tissue interaction using a near infra-red source. Layer-wise spatial distribution of light fluence in the skin tissue is simulated and studied with varying optical properties, indicating the physiological status. This model is expected to help the tissue analysis for non-invasive diagnostics / treatment monitoring.

Keywords: Light-tissue interaction, diffuse optical spectroscopy, Diffusion equation, fluence distribution

Introduction

Soft tissues are longest tissue structures found throughout a human body. It plays a vital role in connecting, providing support or surround other structures and organs of the body. Muscles, skin, tendons, ligaments, nerves, fibrous tissues, fat, blood vessels are commonly classified as soft tissues[1-2].

Soft tissue sarcoma is the most primitive stage of cancer. They originate in mesenchymal cells which are the elementary cells that mature into different types of soft tissues and later develop into cancerous tissue like blood vessel tumors, nerve tissue tumors, liposarcoma etc. [1]. For determining the extent of these diseases different invasive modes like tissue biopsy and blood analysis are commonly employed. Also, there are numerous diagnostic methods like CT, MRI, ultrasound etc. for assessing different types of sarcomas, but they are only able to assess at a later phase of its progression and all these modalities share its own disadvantages [3].

Optical methods present several interesting features for the purpose of medical imaging and its diagnosis[4]. Diffuse optics, where light is operated in the visible and near-infrared (NIR) ranges is nonionizing. Thus, it does not harm biological tissues, provided if the procedures are carried out within the stipulated exposure limits by the concerned agencies. Therefore provides an immense opportunity for evaluating and assessing the extent of the disease while comparing with normal tissue.

Diffuse Optical Spectroscopy is becoming as a front-runner in optical medical imaging as it operates light in the visible or Near Infra-Red (NIR) for analysing the tissue samples and provide vital information about the pathological state of the tissue by evaluating the transmitted or reflected light intensity using multiple source-detector pairs [4–7]. Therefore it opens up an avenue to make use of this technique in determining physiological conditions related to soft tissues like in skin tissue or in ligaments, tendons where invasive methods are required at present. In order to understand the interaction of light on the tissue, it is essential to understand the propagation of light through tissues. Light propagation through tissues is explained using Radiative Transport Equation (RTE) [8].

Radiative transport equation essentially gives total energy contained in a volume element of a scattering medium when light propagates through it. Since RTE is complex and is dependent on a number of variables, a much more simplified version of RTE called as diffusion equation (DE) is used for the analysis of light propagation through tissues [9]. But solving the diffusion equation analytically is computationally expensive and is tedious. Research in this direction has made use of Monte Carlo simulations to determine the fluence. But studies have shown that COMSOL is a more powerful tool for producing more accurate and faster computation than standard Monte Carlo method of light transport in tissues [10]. So a numerical simulation study in this direction was carried out using COMSOL Multiphysics for determining fluence along soft

tissue which can pave way for analyzing and determining various condition affecting these tissues using diffuse optical techniques.

Principle

Diffusion equation is primarily used for studying the propagation of light in biological tissues [11]. Figure 1 shows the schematic of interaction of light inside a tissue volume and its different modes of propagation within the tissue. Optical properties like absorption coefficient and scattering coefficient are the prime factors that contribute to the change in the reflected or transmitted intensity obtained at the detector. P1 approximation of radiative transport equation is generally referred to as the diffusion equation (DE) and is given by [8]

$$\frac{\partial u(\vec{r},t)}{v\,\partial t} + \mu_a\, u(\vec{r},t) - c\nabla^2 u(\vec{r},t) = I(\vec{r},t) \qquad (1)$$

Where r denotes position vector of the photon, t is the instantaneous time at which source is applied, v is the speed of light in the medium, u is the photon fluence rate is defined as the energy flow per unit area per unit time in the given medium, I is intensity of the source provided and c is the photon diffusion coefficient where

$$c = \frac{1}{3(\mu_a + \mu_s')} \qquad (2)$$

Where, μ_a is the absorption coefficient and μ_s' is the reduced scattering coefficient of the medium.

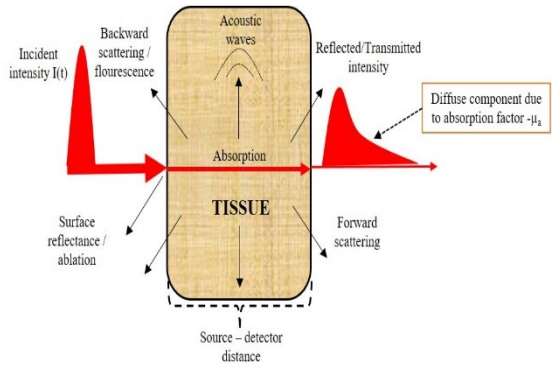

Figure 1: Representation of different modes of light interaction within a tissue.

Due to the scattering effect, the light gets dispersed in numerous direction within the tissue and the peak in the detected intensity in figure 1 represents the scattering component. Whereas the exponential decay in the detected intensity is due to the contribution of absorbers like melanosomes, fats etc. and the intensity from the incoming source is attenuated due to the presence of these absorbers.

Some of the essential conditions for the DE as given by (1) to be valid are [8] that the reduced scattering coefficient should be much greater than the absorption coefficient so that the light radiance is approximately isotropic throughout the volume. Also, the source must be isotropic and the photon flux should not vary much in time.

So DE is usable in biological tissues for studying light propagation as it satisfies conditions mentioned above. However, the radiance ceases to be isotropic near a boundary (such as the one between the dermis and epidermis in a normal human skin) and hence, suitable boundary conditions must be applied on DE equation.

Boundary condition

The partial flux boundary condition (also known as Robin boundary condition) that relates the fluence rate to its gradient at the boundary is calculated by substituting the value of radiance ($\hat{\Omega}$) from DE and an appropriate form of Fresnel reflection coefficient in equation [11]:

$$u = z_b \hat{n} \cdot \nabla u \qquad (3)$$

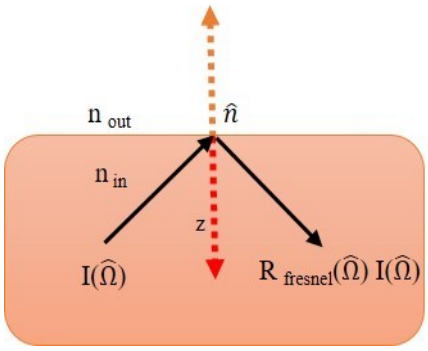

Figure 2: Tissue cross section showing the effect of boundary condition applied to it

Equation (3) gives the fluence rate on the boundary and z_b is the z-coordinate of the boundary.

$$z_b = \frac{2l_{tr}(1+R_{eff})}{3(1+R_{eff})} \qquad (4)$$

Where l_{tr} is the mean free path length, R_{eff} is the effective reflection coefficient which accounts for the index mismatch between the diffusing medium and the air. Figure 2 represents the effect of boundary condition in a tissue cross-section. The

value of R_{eff} in the equation ranges from 1 – 2.2 for soft tissues [12-13].

Calculation of optical properties of skin

To carry out a simulation of light propagation in skin tissue, information about optical properties like absorption coefficient and reduced scattering coefficient of different skin layers is required. Figure 3 shows the schematic of different layers of a human skin tissue with their mean thickness and the effect of light interaction with these tissue. The reflected light intensity decreases as depth increases. Numerous works for its estimation have been published earlier [9], [14–16]. The major absorber in the epidermal layer is contributed by melanosomes. Concentration of melanosomes in epidermal layer vary from 10% - 40% for humans [13]. The presence of melanosomes and its concentration is the reason for varying skin tones for humans which are classified according to the Fitzpatrick scale.

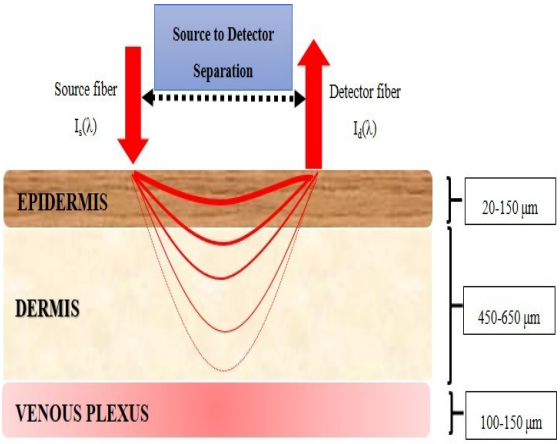

Figure 3: Schematic of Human soft tissue layers and light tissue interaction at different layers.

The absorption coefficient of the dermis is determined by the absorption coefficient of the skin baseline as well as the absorption coefficient of hemoglobin present in the dermal layer. The presence of blood in the dermal layer is only about 0.2% for normal humans. Scattering is due to the contribution of various tissue structures and components like collagen fibers present in these layers. Values for each of these layers when illuminated with a source at a wavelength of 660nm were determined using [14] for a skin tissue having 10 % concentration of melanosomes in epidermal layer and was quantified as shown in table 1.

Table 1: Quantification of optical properties

OPTICAL PARAMETERS	EPIDERMIS	DERMIS
ABSORPTION COEFFECIENT μ_a (cm^{-1})	27.196	0.288
SCATTERING COEFFECIENT μ_s' (cm^{-1})	22.452	22.335
DIFFUSION COEFFECIENT D (cm)	0.00671	0.0147

Simulation study

Light propagation study in a skin tissue model was studied by making use of Helmholtz equation module in COMSOL Multiphysics 5.2a as it replicates the DE explained in the earlier section.

The general form of the Helmholtz equation in COMSOL is given by:

$$\nabla(-c\nabla u) + au = I \quad (5)$$

Where u represents the fluence rate which is the dependent variable. Comparing with the diffusion equation given by (1), diffusion coefficient is given by c, absorption coefficient μ_a given by 'a' and the source term 'I', applied on the tissue slab. Since we are carrying out a stationary study (time independent state), the first term in (1) gets eliminated.

A rectangular block of dimension 10mm × 10mm × 1.1mm, replicating a normal human skin tissue was constructed using COMSOL kernel as shown in Figure 4. The whole block was divided into 2 layers representing an epidermal layer of 0.1mm and dermal layer of 1cm thickness.

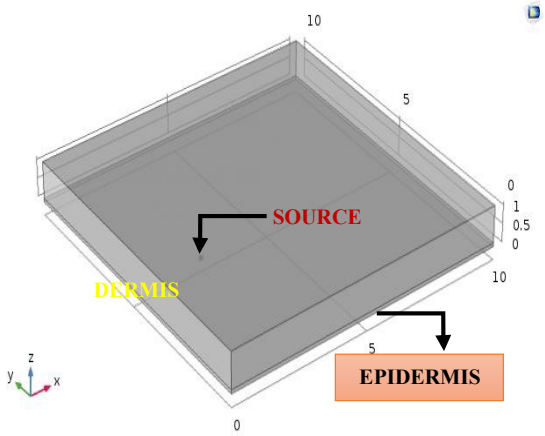

Figure 4: 3D geometry constructed replicating skin tissue

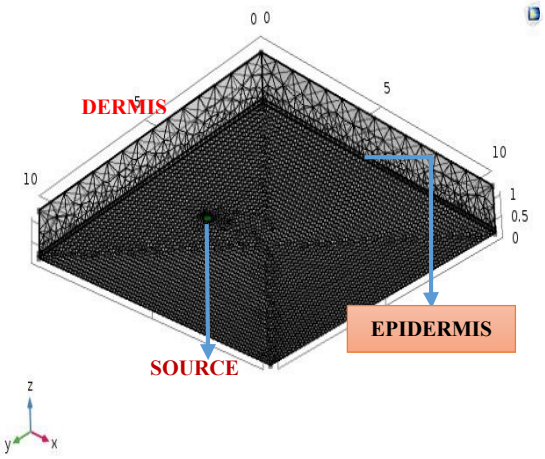

Figure 5: 3D meshed structure

Since low power laser sources are used in diffuse optics methods, a source of 100mW was placed at the epidermal surface of the tissue to carry out the study. This replicates the positioning of laser source on the tissue surface. Helmholtz Physics as given by (5) and necessary input arguments as shown in Table 1 were assigned to each of these layers. Flux source boundary condition which is equivalent to the partial flux boundary condition mentioned in the earlier section was applied to each of these layers. Refractive index values of 1.16 and 1.42 was applied to air - epidermis, and epidermis - dermis boundaries respectively [13] along the boundary condition. A fine meshing was applied to the tissue block and coarse meshing was applied at source point as shown in figure 5.

Results and Discussions

Fluence distribution within the tissue slab was carried out using stationary study and analysis was carried out using 3D slice plots and 2D line plots. Figure 6 shows the 3D sectional views of distribution of light within the tissue block. Figure 7 shows the fluence distribution along the block at different layers. The peak is obtained at the point where the source is placed and the intensity keeps on decreasing as it moves away from the source point. The blue line in the plot represents the net reflected intensity obtained on the surface of tissue block and the red line represents the transmitted intensity at other end of the tissue block. The green light represents the reflected intensity recorded at the epidermal - dermal boundary. The intensity keeps on decreasing as light travels deep inside the tissue layers which shows the contribution of absorbers and its concentration in each of these layers.

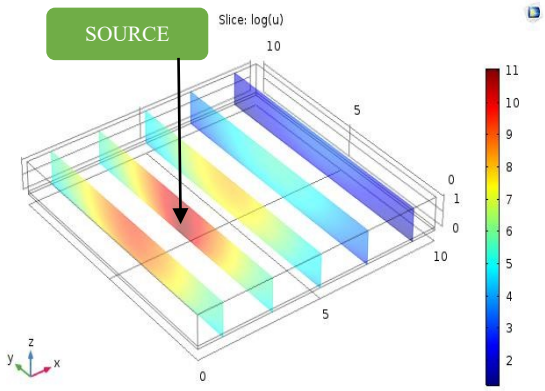

Figure 6: 3D fluence distribution along the tissue block.

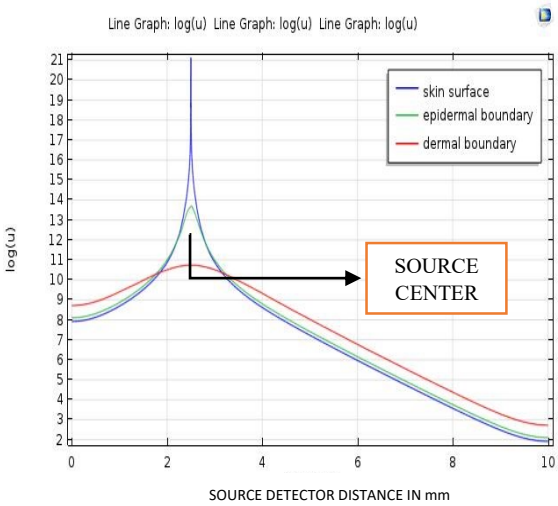

Figure 7: Fluence distribution along the tissue block in different layers

By altering the optical properties of different layers inside the tissue, the distribution of fluence at each of these layers can be studied. Figure 8 shows the variation in fluence at the outer surface on varying the optical properties of the skin. Here the concentration of melanosomes have been varied from 10% - 40% and the reflected intensity obtained along the skin surface have been plotted in figure 8. The reflected intensity determined along the skin surface shows a negative trend as the concentration in the epidermal layer was increased and it was observed throughout the width of the tissue block. A detector can be placed at a definite distance based on the application from the source position and thus the variation in reflected intensity at detector point can be determined. The presence of any pathological condition lead to variation in the reflected intensity and thus by measuring the intensity the extent of its condition can be quantified.

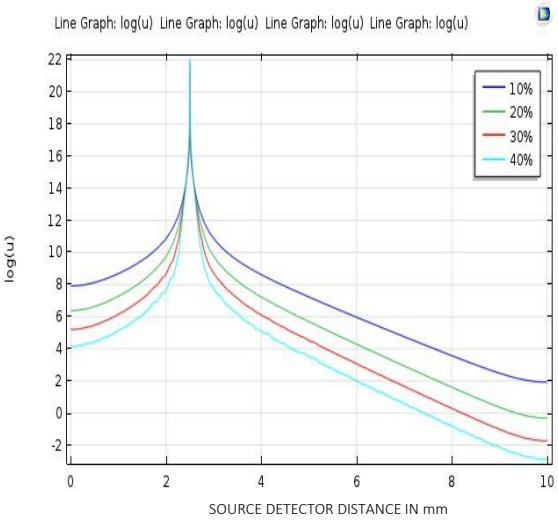

Figure 8: Fluence distribution along epidermal layer for varying optical properties.

Conclusion

Fluence distribution along multiple skin layers was analyzed using COMSOL Multiphysics. The Helmholtz equation module was made use to replicate the diffusion equation which is primarily applied for analysis of light distribution within a tissue. The analysis showed that with variation in optical properties, the fluence distribution at a fixed source to target distance varies and it can be determined using COMSOL with relative ease when compared to previous methods. This could pave way for carrying out simulation studies related to soft tissue abnormalities using diffuse optical techniques and can contribute to its hardware realization.

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