CHAPTER 4

PHANTOM OPTICAL PROPERTY EXTRACTION

The Near Infra Red light incident on the tissue and reflected from the surface of the tissue comprises of two components namely specularly reflected and diffusely reflected components. While the intensity of specular component is largely determined by surface properties of the tissue, the intensity of the diffuse component includes the contribution of the absorbance and scattering light from the tissue substrate.

The biological tissue is an in-homogenous absorbing medium with strong scattering properties. Its top most layers have a specular reflection of light with 5% to 7% of the total incident light. About 93% to 95% of the light is scattered and absorbed by the remaining layers.

4.1 ABSORPTION AND SCATTERING COEFFICIENTS

4.1.1 Absorption Co-efficient

The collimated beam of monochromatic light intensity $I_0$ at wavelength ($\lambda$) 780nm and 850nm illuminates the tissue surface of thickness ‘d’. Therefore the transmitted light intensity is $I_d$, $dl$ is the differential change in intensity and $Idx$ is the intensity of collimated beam light along a small path in a homogenous medium.

$$\int dl = -\mu_a Idx$$  \hspace{1cm} (4.1)

The Lambert-Beer law relates (Yaroslaysky et al. 2002) the absorption coefficient as $\mu_a$ that is absorption coefficient of medium (mm$^{-1}$).
for a given wavelength; it is the probability per unit length of photon being absorbed.

\[ I_d = I_0 e^{-\mu_a d} \]  

Absorption coefficient (Judith et al. 1997) can also be defined as the product of particle density \( \rho \) and absorption cross sectional area \( \sigma_a \). \( 1/ \mu_a \) is the absorption path length, \( \mu_a \) (Subhadra et al. 2003) can be expressed in terms of specific extinction coefficient \( \varepsilon \) [molar\(^{-1}\) mm\(^{-1}\)] and concentration of compound solution \( C \) [molar].

\[ \mu_a = \rho \sigma_a \]  

4.1.2 Scattering Coefficient

In macroscopic scale, the refractive index mismatch between the skin and skull make rise to refraction. The light scattering is caused by the mismatches at the microscopic boundaries on account of cell organelles and cell. The transmission ‘T’ is defined as the ratio of transmitted \( I_d \) to incident intensity \( I_0 \).

\[ T = I_d / I_0 \]  

The optical density or attenuation of the medium describes as

\[ OD = \log_{10} 1/T = -\log_{10} I_d / I_0 \]  

In isotropic membranes, the collimated beam of light travels a mean free path. Similar to absorption coefficient, the scattering coefficient for a collimated source using Lambert-Beer law is given by

\[ I_d = I_0 e^{-\mu_s d} \]  

Scattering coefficient (Tuchin 1997) is also expressed in terms of particle density \( \rho \) and scattering cross sectional area \( \sigma_s \).
\[ \mu_s = \rho \sigma, \quad (4.7) \]

The scattering path length \(1/\mu_s\) is the average distance where a photon travels between consecutive scattering events. Optical thickness is \(\mu_s d\) in terms of scattering lengths.

The scatter rays has a phase function \(p(\theta)\) expressed as cosine function of scattering angle commonly known as Henyey–Greenstein (HG) phase function

\[ p(\theta) = p(\cos \theta) \quad (4.8) \]

Mie scattering (Gebhart et al. 2006) theory was developed that analytical solutions of the phase function have a limit since the size of the scatter is less than wavelength \(\lambda\) of incident photon. Mie can be approximated by Rayleigh’s theory of scattering. The anisotropy ‘g’ (Wai-Fung Cheong et al. 1990) is mean cosine of scattering angle.

\[ g = \frac{1}{\pi} \int_{-1}^{1} \cos \theta p(\cos \theta) d(\cos \theta) \quad (4.9) \]

For \(g=0\) it is perfectly isotropic scattering, \(g=1\) complete forward scattering of incident wave occurs. In case of soft tissue the anisotropy factor ‘g’ is specified to be in the range \(0.7 \leq g \leq 0.9\).

The transport or reduced scattering coefficient \(\mu_s'\)

\[ \mu_s' = \mu_s(1 - g) \quad (4.10) \]

Mean path travelled by means of a collimated beam of light in an isotropic medium is \(1/\mu_s'\), where as \(1/ \mu_t\) is the mean free path between absorption and scattering.

Total attenuation coefficient \(\mu_t\)

\[ \mu_t = \mu_s + \mu_a \quad (4.11) \]
Transport attenuation coefficient is given below

$$\mu_{tr} = \mu_a + \mu_s (1 - g) = \mu_a + \mu_s^t$$  \hspace{1cm} (4.12)\\

(i) **Rayleigh Scattering**

Rayleigh scattering depends upon the relative refractive index $n_r$ and the size of the tissue $x$ considered as spherical structure for scattering radiation.

$$x = 2\pi a / (\lambda / n_{med})$$  \hspace{1cm} (4.13)

$a << 1$, radius of the sphere, $\lambda$ is wavelength of NIR light, $n_{med}$ is the turbid medium refractive index of tissue.

$$n_r = n_p / n_{med}$$  \hspace{1cm} (4.14)

$n_p$ is the refractive index of air. In Rayleigh scattering, dipole moment of particle $p = \alpha E$, i.e. $p$ is proportional to instantaneous Electric Field (EF) vector. $\alpha$ is scalar for an isotropic spherical particle. From energy of EF produced by oscillating dipole, the intensity of scattered radiation

$$I_s = (1 + \cos^2 \theta) I_0 \frac{N k^2 \alpha^2}{2R^2}$$

where wave number $k = 2\pi / \lambda$, $\alpha \approx r \frac{1}{n_r^2 - 1}. The intensity of Rayleigh scattering $I_s$ is given by

$$I_s = \frac{8\pi^4 N r^6}{\lambda^2 R^2} \left| \frac{n_r^2 - 1}{n_r^2 + 2} \right|^2 (1 + \cos^2 \theta) I_0$$  \hspace{1cm} (4.15)

$N$ is the number of scatters, $r$ is particle size in polarization factor, $R$ is the distance from scatter and $I_0$ is the incident intensity. Rayleigh scattering (Dorota et al. 2009) takes place at a depth of 0.1µm to 0.01µm in soft tissue. At $\theta = 90^\circ$, Rayleigh scattering is half of the forward intensity and proportional to $\lambda^{-4}$. Rayleigh scattering phase angle is $p_{rayleigh}(\theta)$

$$p_{rayleigh}(\theta) = 3/4 (1 + \cos^2 \theta)$$  \hspace{1cm} (4.16)
(ii) Mie Scattering

Mie scattering (Jacques et al. 2008) in tissue happens within the micro molecular region from 1\(\mu\)m to 0.1\(\mu\)m. Mie scatters also depends on size \(x\) and relative refractive index \(n_r\), with its intensity proportional to \(\lambda^{-2}\) (Judith et al. 1997).

The anisotropy function \(g\) (Flock et al. 1992) is determined by the phase function of radiance \(L(r, \hat{s}, t)\).

\[
g = \frac{\int_0^\pi p_{\text{mic}}(\theta) \cos(\theta) \cdot 2\pi \sin \theta d\theta}{\int_0^\pi p_{\text{mic}}(\theta) \cdot 2\pi \sin \theta d\theta}
\]

(4.17)

Reduced scattering coefficient \(\mu_s\) (Steven 2013) combines Mie and Rayleigh scattering which is defined as

\[
\mu_s'(\lambda) = a \left(\frac{\lambda}{500(\text{nm})}\right)^{-4}
\]

(4.18)

Scaling factor ‘a’ has value \(\mu_s'(500\text{nm})\). The reduced scattering coefficient parameter (Steven 2013) distinguishes the layers as epidermis, dermis, skull, gray matter and white matter.

\[
\mu_s'(\lambda) = a \left(f_{\text{ray}} \left(\frac{\lambda}{500(\text{nm})}\right)^{-4} + (1-f_{\text{ray}}) \left(\frac{\lambda}{500(\text{nm})}\right)^{-b_{\text{mic}}}\right)
\]

(4.19)

where \(f_{\text{ray}} [0.3-0.82]\) is a fraction of Rayleigh scattering \(p_{\text{rayleigh}}\) and \(b_{\text{mic}} [1.1-3.2]\) is the Mie scattering power, whose range for soft tissues namely brain tissue is as mentioned.

4.2 DIFFUSION APPROXIMATION

The Boltzmann transport equation defines the photon propagation (Fishkin & Gratton 1993) from source towards the tissue surface as
\[
\frac{1}{c} \frac{\partial L(r, \hat{s}, t)}{\partial t} + \nabla L(r, \hat{s}, t) \hat{s} = - (\sigma + \beta) L(r, \hat{s}, t) + \sigma \int_{4\pi} L(r, \hat{s}, t) f(\hat{s}, \hat{s}') d\Omega' + Q(r, \hat{s}, t)
\]

\[(4.20)\]

\(L(r, \hat{s}, t)\) represents radiance in W/(m²sr), \(\hat{s}\) unit vector pointing in desired direction, \(\sigma = 1/\mu_a\) (mm), \(\beta = 1/\mu_s\) (mm). The normalized differential scattering cross section \(\int_{4\pi} f(\hat{s}, \hat{s}') d\Omega' = 1\), \(Q(r, \hat{s}, t)\) is the source power injected in unit solid angle centered on \(\hat{s}\) in a unit volume at \(\cdot r\), \(c' = 3 \times 10^8\) m/s that is the velocity of light.

The fluence rate \(\phi\) and the flux \(j\) defined by the integration over solid angle (Uma Maheswari & Sathiyamoorthy 2016) are obtained as

\[
\frac{1}{c} \frac{\partial \phi(r, t)}{\partial t} + \nabla j(r, t) = - \beta \phi(r, t) + S(r, t)
\]

\[(4.21)\]

\[S(r, t) = \int_{4\pi} Q(r, \hat{s}, t) d\Omega\]

\[(4.22)\]

\[\phi(r, t) = \int_{4\pi} L(r, \hat{s}, t) d\Omega\]

\[(4.23)\]

\[j(r, t) = \int_{4\pi} L(r, \hat{s}, t) \hat{s} d\Omega\]

\[(4.24)\]

The diffusion equation is derived as follows

\[
D \nabla^2 \phi(r, t) - \mu_a \phi(r, t) = \frac{1}{c} \frac{\partial \phi(r, t)}{\partial t} - S(r, t)
\]

\[(4.25)\]

\(D\) is the diffusion coefficient, \(D = 1/ (\mu_s + \mu_a)\). The diffusion equation solution was carried out in infinite, semi-infinite and slab or layered structure of homogeneous medium. They assume an isotropic source and attain the solution for fluence rate (Dmitry & Laurent 2010) at appropriate boundary conditions (Richard et al. 1994).
4.3 BOUNDARY ELEMENT SOLUTION

4.3.1 Infinite Homogenous Medium

The NIR light propagation through a spherical shaped tissue structure in a turbid medium is said to be an infinite homogenous medium. The source \( S(r, t) \) is given in terms of \( \delta(r) \) and \( \delta(t) \) respectively. Figure 4.1 shows the location of source and detector with radius \( r \) in a spherical symmetry.

Diffusion equation (Alwin et al. 1998) is a partial differential equation and the solution using Green’s function is

\[
\phi(r,t) = c' (4\pi Dc' t)^{-3/2} \exp(-\mu_d c' r) \exp(-r^2 / 4Dc' t) \quad (4.26)
\]

where \( c' = c / n \), ‘n’ is the refractive index of the tissue, ‘r’ is the radial distance from source.

![Figure 4.1 Infinite Homogenous Geometry](image)

4.3.2 Semi-infinite Medium

In semi-infinite medium, the tissue exhibits cylindrical symmetry structure with half space geometries (Alwin & Michael 1997) and the location of the source and detector in cylindrical co-ordinates are as shown in Figure
4.2. The medium (Alwin & Michael 1994) is along the positive z axis, (photon pulse) while the x-y plane acts as a boundary. \( z_0 \) is termed as the mean free path length of the medium.

\[
z = z_0 = 1/\left(\mu_a + \mu_s\right)
\]  

(4.27)

The Fresnel reflection occurs at the boundary surface of the tissue and the fluence rate \( \phi \) at the boundary (on surface of turbid medium) is zero, so it is called as Zero Boundary Condition (ZBC). Beyond the boundary, \( \phi \) is not exactly zero but an extrapolated boundary at a distance \( z_e \) exits outside the real boundary, so we can define it as Extrapolated Boundary Condition (EBC). \( z_e \) is the function of the difference between two media. Hence, for tissue-air boundary is \( z_e \cong 2z_0 \).

**Figure 4.2 Semi-infinite Geometry**

The diffusion equation solution using two boundary conditions namely ZBC and EBC determines the fluence rate in the medium as

\[
z_b = \frac{1 + R_{\text{eff}}}{1 - R_{\text{eff}}} 2D, \quad R_{\text{eff}} \quad \text{(Kienle & Patterson 1996)}
\]

that is the fraction of photons
which internally diffuses reflections at the boundary, \( R_{\text{eff}} = 0.493 \) when refractive index \( n=1.4 \).

\[
\phi(\rho, z, t) = \frac{c'}{(4\pi Dc't)^{3/2}} \exp(-\mu c' t) \exp \left[ -\frac{(z-z_0)^2 + \rho^2}{4Dc' t} \right] - \exp \left[ -\frac{(z + z_0 + 2z_e)^2 + \rho^2}{4Dc' t} \right] 
\]

(4.28)

### 4.3.3 Layered Medium

The layered tissue structure (Juliette et al. 2014, Alwin et al. 1998) has slab geometry as shown in Figure 4.3 with the physical boundary at \( z=0 \) and the source placed at \( z=z_0 \). The extrapolated boundary (Hielscher et al. 1995) is at \( z=-z_e \), where negative source mirror image is on the positive axis and the source ensures zero fluence rate. The negative source is found at \( z_{p0} = -z_0 - 2z_e \), where
z_e = 2D/\eta, \eta = (\gamma \pi)^{-1.5} and \gamma = 4Dc'. The extrapolated boundary condition is given by 
\[ z_0 = 2(d + 2z_e) + z_{p0}, \quad Z_p = 2(d + 2z_e) + z_{p0} \pm 1/\mu_s, \]
where ‘d’ is the slab thickness. The source and mirror image source is at a radial distance 
\[ r_0 = r^2 + (z_0 - z)^2 \quad \text{and} \quad r_p = r^2 + (z_p + z_0)^2, \]
\(\Phi\) is initial value. The fluence rate \(\phi\) in homogenous slab (Patterson et al 1989) structure using Green’s solution is given as

\[
\phi(r, z, t) = c'(\eta t)^{1.5} e^{-\mu'_a c't} \left[ \Phi + \exp(-\eta_0/\gamma t) - \exp(-r_p/\gamma t) \right] \quad (4.29)
\]

A dipole is formed by the original source, a positive source and an imaginary source a negative source. The soft tissue profile layers are similar to layered structure model. Therefore, the accuracy of reduced scattering coefficient is high. The slab model is examined with multiple laser sources namely six and their mirror images are also activated to have a bounded accuracy.

### 4.4 DIFFUSE REFLECTANCE

The reflected rays from the tissue boundary surface create a rise to diffuse reflectance. It is the multiple light scattering within the region of interest. Diffuse Reflectance is the photon current leaving the tissue at \( z=0 \) boundary condition. The diffuse reflectance (Alwin & Michael 1994, Alwin et al. 1996) in case of the semi-infinite turbid medium is determined as the current across the boundary (Farrell et al. 1992).

\[
R_f(\rho, t) = -D\nabla \phi(\rho, z, t).(-z)_{z=0} \quad (4.30)
\]

Substitute equation (4.28) in equation (4.30), \( R_f \) generates (Erik et al. 2008) a spatially dependent diffuse reflectance for an extended scattered source.

\[
R_f(\rho, t) = \frac{1}{2(4\pi Dc')}^{3/2} t^{-5/2} \exp(-\mu_a c't), \left( z_0 \exp\left(\frac{-r_1^2}{4Dc't}\right) + (z_0 + 2z_b) \exp\left(\frac{-r_2^2}{4Dc't}\right) \right) \quad (4.31)
\]
The diffuse reflectance equation has $r_1$ and $r_2$ to be expressed in the form of $r_1^2 = z_0^2 + \rho^2$, $r_2^2 = (z_0 + 2z_b)^2 + \rho^2$. The figure 4.4 represents the flow Chart for Optical Property extraction using three structural medium in BEM.

4.5 SIMULATION RESULT AND DISCUSSION

The response of the photo-detector electrical voltage response is measured and fed to Analog to Digital Converter, then it is further fed to personal computer in the range of (0-5) volts, i.e. (2.63-4.2) volts for normal cells and (4.3-5) volts in case of carcinoma cells.
Table 4.1 Photodiode Response voltages, Absorption and Scattering coefficients with Optical flux

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<th>S. No</th>
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<th>Ph₂</th>
<th>Ph₃</th>
<th>Ph₄</th>
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The tissue structure properties such as diameter and area, along with length of penetration are applied to compute the absorption and scattering coefficient. The length of penetration was evaluated using the photo-detector response voltages. Diffuse Reflectance, attenuation and photonic flux or optical flux were determined by boundary element condition. Table 4.1 presented the photo-detector response voltage along with the estimated absorption coefficient, scattering coefficient and optical flux.

The incident current on the tissue area with output photodiode voltage makes it possible to obtain the power and intensity of the reflected and incident photons. The basic phantom parameters determined are namely Absorbance, Optical Flux Density, Transmittance and Attenuation. Absorption coefficient $\mu_a$ for different fractional volumes of the tissue (0.01-0.21) $\mu$m$^3$ was simulated in MATLAB2013a using the refractive index of the medium $n$ (1.33-1.4). The product of photon particle density with optical flux density OD and extinction coefficient determines the $\mu_a$ incm$^{-1}$. Figure 4.5 depicts the absorption coefficient variation along the NIR wavelength (700-1200) nm for different fractional volumes of the spherical structure of the phantom.

Mie Scattering dominates in tissue since the scattering of light by particles of a size on the same order as the wavelength of light. Mie scattering is sharper for large particles which are responsible for white appearance of adipose tissue in brain.

Figure 4.6 depicts the Anisotropy factor ‘$g$’ average cosine phase function plotted for NIR wavelength versus $g$ ($\theta$) for various tissue diameter. The Mie scattering coefficient is limited by anisotropy factor. The anisotropy factor value $g = [-1 0 1]$, where $g=-1$ for backward, $g=0$ for isotropic and $g=1$
for forward scattering. Optimum value of g from the graph was referred to be in range from 0.7 to 0.9.

**Figure 4.5** Absorption Coefficient $\mu_a$ in cm$^{-1}$

**Figure 4.6** Anisotropy Coefficient ‘g’
Figure 4.7 shows the Mie scattering coefficient plotted for several tissue diameter ranging from [0.1-10] µm. The Mie scattering is predominant if it is a macro particle.

![Mie Scattering Coefficient](image)

**Figure 4.7 Mie Scattering Coefficient $\mu_s$ in cm$^{-1}$**

![Rayleigh Scattering](image)

**Figure 4.8 Rayleigh scattering $\mu_s$ in cm$^{-1}$**
Rayleigh Scatter contributes for interactions with small particles and strength of scattering depends on wavelength. It is also called elastic scattering because the scattered photons have same energy as incident photons.

Figure 4.8 illustrates how Rayleigh scattering Coefficient $f_{\text{ray}}$ was calculated for various diameters of the tissue ranging from [0.1-10] µm.

Reduced scattering coefficient $\mu_s$ combines the Mie and Rayleigh scattering coefficients with their scattering powers. Figure 4.9 shows the Reduced Scattering Coefficient for NIR wavelengths depended on the anisotropy factor $g$. The reduced scattering coefficient was also evaluated for different diameters from 0.1µm to 10µm of tissue thickness. Diffusion of light rays on tissue profile is suitably characterized by the reduced scattering coefficient. As the multiple scattering events occur it was to analyze high scattering coefficient as a complicated process.

![Reduced Scattering Coefficient](image)

**Figure 4.9 Reduced Scattering Coefficient $\mu_s$ in cm$^{-1}$**
As a result, the scattering coefficient is reduced based on anisotropy factor ‘g’. The reduced scattering coefficient value ranges from 4 to 19. From the plot, it illustrates that the reduced scattering has more pronounced effect from 700-900nm wavelengths in NIR.

Figure 4.10 depicts the performance of Attenuation factor \( \mu_t \) owing to \( \mu_a \) and \( \mu_s \). The measured and designed values of attenuation factor were compared and later the error percentage is computed to be 0.25%.

![Attenuation Factor](image)

**Figure 4.10 Attenuation factor**

The diffusion equation solved by zero and extrapolated boundary conditions with three different geometrical structures such as spherical, cylindrical and slab structures of the phantom are analyzed along with placement of source and detectors. The fluence rate \( \phi(r, t) \) was compared for
different geometrical structures with extrapolated boundary conditions in boundary element method.

![Boundary conditions](image)

**Figure 4.11 Boundary Conditions**

Figure 4.11 depicts the fluence rate at refractive index $n=1.33$, the depth the light travels in tissue $z=0.008\text{m}$, the radius of the tissue $r=0.008\text{m}$ where the fluence rate was calculated in the range of time $t$, 0 to 5 ns. The slab thickness $d=0.01\text{m}$, the photon flux was calculated in between the boundary of slab structure with a turbid medium. Normally the semi-infinite medium is preferred for analysis due to a cylindrical symmetry tissue structure with light propagation in an infinite medium.

In Figure 4.12 Cylindrical symmetric fluence image is obtained in semi-infinite boundary condition with refractive index $n=1.4$, initial detector
depth $z=6\text{mm}$, initial radial range of the detector $r = 8\text{mm}$ and resolution parameter $r_f = 50$. The fluence rate was calculated in three dimensional array containing $z$, $r$ and $t$ (time in nanoseconds).

**Figure 4.12 Cylindrical symmetric fluence**

**Diffuse Reflectance**

**Figure 4.13 Diffuse Reflectance in logarithmic scale**
Figure 4.13 Diffuse Reflectance was simulated in logarithmic scale where the curve is exponential of the Optical-flux Density (OD) along with effective penetration coefficient $\delta$ and absorption coefficient $\mu_a$. The effective penetration depth $\delta$ of the NIR light in brain tissue is calculated as 1.4cm.

Figure 4.14 portrays the Diffuse Reflectance image of the phantom due to diffuse light scattering in all directions. Irregular phantom has the reflectance rays to be reflected at greater angles and wavelengths. The slide bars adjust the color (0-100) % of the reflected light, where the highest reflection determines tumor phantom and the uniform color determines the normal tissue surface.

**Diffuse Reflectance**

![Image of Diffuse Reflectance Model]

**Figure 4.14 Diffuse Reflectance Model**

Reflectance calculated outside the boundary of the geometrical structure is diffuse reflectance. Figure 4.15 shows the Diffuse Reflectance plot in time domain, defining the reflections outside the slab of thickness $d=5$cm with various radius of reflectance $r= [0.1 0.5 1 5 10]$ cm.
The comparison of the optical parameters from normal tissue cell to tumor tissue cell was shown vividly in Figure 4.16 Absorption Coefficient, Figure 4.17 Scattering Coefficient and Figure 4.18 Diffuse Reflectance. Figure 4.16 illustrates Absorption coefficient $\mu_a$ for normal brain tissue ranges [0.5-0.9] cm$^{-1}$, however in case of tumor tissue range [1.1-1.7] cm$^{-1}$.
Figure 4.17 Scattering Coefficient

Figure 4.17 depicts the scattering coefficient ranges [30-70] cm\(^{-1}\) for normal tissue whereas in scattering tumor tissue [90-200] cm\(^{-1}\). Simulation for the absorption and scattering coefficients is carried out for NIR wavelengths (700-1200) nm.

The Diffuse Reflectance in time domain for brain tumor tissue occurs within 2.5ns faster than the normal tissue. Scattering or reflection is more pronounced in tumor cell compared to normal cell. Figure 4.18 demonstrates the photon propagation through soft tissue in time domain. The time domain reflectance simulated was normalized and measured by an arbitrary unit (a.u.).

The standard values of absorption coefficient and scattering coefficient for 780nm and 850nm are \(\mu_a=0.034, 0.045\) then \(\mu_s=9.3,10\). The Optical properties were compared with previous paper results and we had found that the scattering parameter decays as the wavelength decreases.
Figure 4.18 Diffuse Reflectance

Table 4.2 Optical parameters compared with the Optical Property Extraction System displays the error percentage of the system. The scattering coefficient error is reduced to 0.1% for 780nm and 0.2% for 850nm wavelengths. The absorption coefficient is associated with error percentage of 0.58 for 780nm and 0.2 for 850nm wavelengths.

In system (Sultana et al. 2002), the error percentage of scattering coefficient was found to be larger as 24% compared to absorption coefficient with 1.2%. Here the system (Min–Gyu Choi 2013) appearing with large error percentage of 24.7% and 11% System (Ilias et al. 2010) has a nominal error percentage of 4.3% and 4% which is within the acceptable limit. System (Eiji & David 2003) and system (Steen 2013) have any one parameter error percentage to be high.
### Table 4.2 Optical Property Comparison with Optical property Extraction System

<table>
<thead>
<tr>
<th>System</th>
<th>Optical Parameters (cm(^{-1}))</th>
<th>780nm</th>
<th>850nm</th>
<th>G</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Measured (cm(^{-1}))</td>
<td>Error</td>
<td>Measured (cm(^{-1}))</td>
<td>Error</td>
</tr>
<tr>
<td>Sultana et al 2002</td>
<td>11.57</td>
<td>0.24</td>
<td>10.120</td>
<td>0.012</td>
</tr>
<tr>
<td></td>
<td>0.036</td>
<td>0.05</td>
<td>0.0438</td>
<td>0.026</td>
</tr>
<tr>
<td>Ilias et al 2010</td>
<td>9.700</td>
<td>0.04</td>
<td>9.6000</td>
<td>0.040</td>
</tr>
<tr>
<td></td>
<td>0.107</td>
<td>2.10</td>
<td>0.1060</td>
<td>1.350</td>
</tr>
<tr>
<td>Min–Gyu Choi 2013</td>
<td>11.60</td>
<td>0.24</td>
<td>11.100</td>
<td>0.110</td>
</tr>
<tr>
<td></td>
<td>0.170</td>
<td>4.00</td>
<td>0.1860</td>
<td>3.130</td>
</tr>
<tr>
<td>Eiji &amp; David 2003</td>
<td>9.100</td>
<td>0.21</td>
<td>7.5000</td>
<td>0.250</td>
</tr>
<tr>
<td></td>
<td>0.140</td>
<td>3.11</td>
<td>0.0470</td>
<td>0.040</td>
</tr>
<tr>
<td>Steen 2013</td>
<td>7.5000</td>
<td>0.250</td>
<td>0.0470</td>
<td>0.040</td>
</tr>
<tr>
<td></td>
<td>0.0451</td>
<td>0.002</td>
<td>0.0451</td>
<td>0.002</td>
</tr>
<tr>
<td>Optical property Extraction System</td>
<td>9.2900</td>
<td>0.001</td>
<td>10.020</td>
<td>0.002</td>
</tr>
<tr>
<td></td>
<td>0.0342</td>
<td>0.005</td>
<td>0.0451</td>
<td>0.002</td>
</tr>
</tbody>
</table>

Figure 4.19 shows the error percentage at 780nm wavelength (Sultana et al. 2002, Ilias et al. 2010, Min–Gyu Choi 2013, Eiji & David 2003) for anisotropy factor g=0.7 results obtained are minimum (0.1% or 0.2%) for a non invasive optical parameter extraction system. Error calculated with respect to standard values for VCSEL laser diodes OPV310 and D7805I is least due to the monochromatic beam of red light which is not harmful, compared to blue light.
The avalanche photodiode OPT101 is insensitive to noise voltages on account of shot noise or thermal noise and therefore the error measured, is negligible at 850nm NIR wavelength.

The most preferred NIR Wavelength range for optical tomography with reduced error ranges from 700nm to 890nm. This NIR range is the best range for non invasive techniques with high penetration depth of (2-3) cm and no side effect for the patients. Figure 4.20 shows the comparison of optical wavelength at 850nm (Sultana et al. 2002, Ilias et al. 2010, Min–Gyu Choi 2013, Steen 2013).
CLINICAL RESULT ANALYSIS

4.6

The soft tissue sarcoma patients are diagnosed with imaging methods namely CT injected with dye which assists in clear visuals of the organ images. The MRI using radio waves, generated with the help of magnetic field interaction projects a series of detailed pictures on the affected areas. The PET imaging injects a radio isotope for intense imaging discrimination of the tumor from normal cells. After screening test’s confirmation of tumor presence, the tumor is subjected to biopsy, to determine the stage of soft tissue sarcoma. The initial treatment is chemotherapy or radiation therapy and later on, the soft tissue sarcoma is staged again, based on the differentiation, mitotic count and necrosis.

The DOT imaging method is enhanced to avoid the side effect for patients due to other imaging method. The DOT screening technique is non-invasive, non-ionizing and cost effective portable device for predicting the
presence of soft tissue tumor. Instead of biopsy, from the reconstructed image segmentation, the affected volume of the tumor cells is diagnosed without any incisions on the patient.

The DOT instrument designed to measure the presence of tumor in sarcoma patients before the surgery is as shown in the Table 4.4. The confirmation report using MRI in brain tumor and mammogram in breast tumor is illustrated in the Table 4.3 given below. The subject data were obtained from the following hospitals in South India viz., Harshamitra Super Speciality Cancer Centre and Research Institute, Tiruchirappalli and Radiation Oncology Centre at Kavery Medical Centre, Tiruchirappalli, Tamil Nadu, India (KMC (2007), Harshamitra (2016)). The hospital data were obtained by getting the consent of the patient, through the research supervisor.

**Table 4.3 Patient Report for Sarcoma Detection**

<table>
<thead>
<tr>
<th>Patient ID/Date of Detection</th>
<th>Age</th>
<th>Disease</th>
<th>MRI/Mammogram Result</th>
</tr>
</thead>
<tbody>
<tr>
<td>Patient 105 (06.01.2013)</td>
<td>52</td>
<td>Brain tumour</td>
<td>MRI scan image predicts soft tumour consistency on the left hemisphere with subtle endema and mass effect on fourth ventricle</td>
</tr>
<tr>
<td>Patient 405 (19.03.2013)</td>
<td>62</td>
<td>Breast Cancer</td>
<td>CA{L}Breast T2/T4bN1M0 ER+, PR+ cErb B2+2+weak position. Biopsy: Ductal CA in situ, Ng High, with small focus S/O Invasion</td>
</tr>
</tbody>
</table>
While the Table 4.3 detects the presence of tumor based on MRI and Mammogram results, the Table 4.4 illustrates the testing and diagnosis using DOT instrument manipulation of absorption and scattering coefficient using the Lambert-Beer law. From the experimental results, the absorption and scattering coefficient were manipulated from which the tumor presence is predictable as per the confirmation of clinical diagnosis screening.

### 4.7 SUMMARY

In this chapter, the result of optical parameter of soft tissue namely $\mu_a$ and $\mu_s$ calculated from measured photo detector OPT101 voltages in 780nm and 850nm. The simulated result of attenuation was calculated and measured with an error 0.25%. The investigation was done on the performance of the system, by varying the position of source and detector, hence predicting the geometrical symmetry of the tissue in turbid medium. The fluence rate $\phi$ measured with arbitrary unit was found to be maximum in semi-infinite boundary with refractive index $n=1.4$, range of radial distance of detector
r=8mm, depth of penetration as z=6mm and resolution parameter $r_f=50$. The Diffuse reflectance in time domain and frequency domain were evaluated and their performance was examined with the high reflection from tumor compared to normal phantom which is depicted in the Figure 4.14 with change of color in frequency domain. The occurrence of reflections from tumor phantom is at a faster rate within 2ns of time compared to normal tissue reflections, occurring within 4ns as shown in the Figure 4.18. The Performance of the system was also analyzed with the error % at 780nm and 850nm wavelength with negligible error measured $\mu_a (780nm) =0.58\%$, $\mu_s' (780nm) =0.1\%$, $\mu_a (850nm) =0.2\%$, $\mu_s' (850nm) =0.2\%$ as shown in the Figure 4.19 and 4.20 by comparison with existing systems. The Clinical result analyses with the help of the patients were also analyzed, to demonstrate the functionality of designed DOT instrument.