CHAPTER 4

DEVELOPMENT OF AN OPTICAL IMAGING SYSTEM
FOR IMAGING THICK TISSUES - PHASE I

4.1 NEED FOR AN OPTICAL IMAGING SYSTEM

Medical imaging techniques currently in use like MRI, CT and Ultra sound are costly and are out of reach of the majority of our population. The need for a low cost but effective imaging technique can be a boon for the deprived majority. Optical imaging presents several potential advantages over existing radiological techniques being nonionic, inexpensive and providing better spatial and temporal resolution.

The factors that limit this optical imaging technology are the scattering property of the biological tissue and the thickness of the tissue. As light travels through a tissue it is scattered by the numerous refractive index variations in the tissue. Scattering plays an important role in limiting the imaging depth, as it decreases the number of excited photons reaching the particular depth. Consequently only few photons are able to reach greater depths without much scattering. Therefore extraction of the weakly scattered light from the highly scattered light can yield images with increased contrast. In this work, the
problem related to scattering of light is solved by polarization imaging and image processing.

The research towards the development of an optical imaging system capable of imaging tissues of thickness in the order of centimetres has been carried out in three phases. The current chapter discusses the first phase of the system development.

4.2 PHYSICS OF SCATTERING AND ABSORPTION

4.2.1 Scattering and Absorption

Scattering of electromagnetic waves by a system is related to the heterogeneity of that system. If an obstacle which could be a single electron, an atom or molecule, a solid or liquid particle is illuminated by an electromagnetic wave; electric charges in the obstacle are set into oscillatory motion by the electric field of the incident wave. The accelerated electric charges radiate electromagnetic energy in all directions. This is the radiation scattered by the obstacle. In addition to reradiating electromagnetic energy, the excited elementary charge may transform part of the incident electromagnetic energy into other forms and this called absorption.
In biomedical optics, absorption and scattering of photons are most important events. Scattering and absorption are used for both spectroscopic and imaging applications.

- Absorption is the primary event that allows a laser or other light source to cause a potentially therapeutic effect on a tissue. Without absorption, there is no energy transfer to the tissue and the tissue is left unaffected by the light.

- Absorption of light provides a diagnostic role such as the spectroscopy of a tissue. Absorption can provide a clue as to the chemical composition of a tissue, and serve as a mechanism of optical contrast during imaging.

- Scattering provides feedback during therapy. For example, during laser coagulation of tissues, the onset of scattering is an observable endpoint that correlates with a desired therapeutic goal. Scattering also strongly affects the dosimetry of light during therapeutic procedures that depend on absorption.

- Scattering has diagnostic value as well. Scattering depends on the structure of a tissue. Scattering measurements are an important diagnostic tool irrespective of whether one measures the wavelength dependence of scattering, the polarization dependence of scattering, the angular dependence of scattering or the scattering of coherent light.
4.2.2 Single scattering

Consider an arbitrary particle that is conceptually subdivided into small regions. An incident electromagnetic wave induces a dipole moment in each region. These dipoles oscillate at the frequency of the applied field and therefore scatter secondary radiation in all directions. At a distant point, the total scattered field is obtained by superposing the scattered wavelets and due account is taken of their phase differences.

In natural environments however, collections of very many particles occur. Particles in collection are electro magnetically coupled. The field scattered by a particle depends on the total field to which it is exposed. Considerable simplification results if single scattering is assumed and the proposed model makes this assumption.

4.2.3 Absorption and scattering coefficient

The absorption coefficient \( \mu_a [cm^{-1}] \) describes a medium containing many chromophores at a concentration described as a volume density \( \rho_a [cm^3] \). The absorption coefficient is essentially the cross-sectional area per unit volume of medium.

\[
\mu_a = \rho_a \sigma_a
\]

[cm^{-1}] [cm^{-3}][cm^2]

(4.1)

and \( \sigma_a [cm^2] \) is called the effective cross-section.
Figure 4.1 Illustration of: (a) absorption; (b) scattering
The scattering coefficient $\mu_s$ [cm$^{-1}$] describes a medium containing many scattering particles at a concentration described as a volume density $\rho_v$ [cm$^3$]. The scattering coefficient is given by

$$\mu_s = \rho_s \sigma_s$$

[cm$^{-1}$] [cm$^{-3}$][cm$^2$] \hspace{1cm} (4.2)

and $\sigma_s$ [cm$^2$] is called the effective cross-section.

4.2.4 Anisotropy and Reduced scattering coefficient

The anisotropy, $g$ [dimensionless], is a measure of the amount of forward direction retained after a single scattering event. Imagine that a photon is scattered by a particle. A scattering event causes a deflection at angle $\theta$ from the original forward trajectory. $\psi$ is the azimuthal angle of scattering and is shown in the Figure 4.2. The component of the new trajectory which is aligned in the forward direction is shown in red as $\cos(\theta)$. The average deflection angle and the mean value of $\cos(\theta)$ is defined as the anisotropy.
Figure 4.2  Illustration of scattering of a photon
The reduced scattering coefficient is a lumped property incorporating the scattering coefficient $\mu_s$ and the anisotropy $g$ and is given by

$$\mu'_s = \mu_s(1 - g) \text{ [cm}^{-1}\text{]}$$  \hspace{1cm} (4.3)

The purpose of $\mu'_s$ is to describe the diffusion of photons in a random walk of step size of $1/\mu'_s$ [cm] where each step involves isotropic scattering.

### 4.2.5 Polarization

In addition to its irradiance and frequency, an electromagnetic wave has a property called its state of polarization. Consider a plane monochromatic wave with angular frequency $\omega$ and wave number $k$, which is propagating in the $z$ direction in a nonabsorbing medium. The electric field vector at any point lies in a plane the normal to which is parallel to the direction of propagation. In a particular plane, say at $z = 0$, the tip of the electric vector traces out a curve.

$$E(z = 0) = A \cos \omega t + B \sin \omega t$$  \hspace{1cm} (4.4)

Equation (4.4) describes an ellipse called the vibration ellipse and the monochromatic wave is said to be elliptically polarized. If $A = 0$ or $B = 0$, the vibration ellipse is just a straight line and the wave is linearly polarized. If $|A| = |B|$ and $A \cdot B = 0$, the vibration ellipse is a circle and the wave is circularly polarized.
A vibration ellipse is characterized by its irradiance, handedness, direction of rotation of the ellipse, ellipticity (ratio of the length of its semiminor axis to its semimajor axis), and its azimuth (the angle between the semimajor axis and an arbitrary reference direction). These are called the ellipsometric parameters of a plane wave.

4.2.6 Stokes Parameters

Although the ellipsometric parameters completely specify a monochromatic wave of a given frequency, they are not conducive to understanding the transformations of polarized light. The Stokes parameters [142] are an equivalent description of polarized light and are particularly useful in scattering problems.

The Stokes vector of a light beam is given by

\[
[I \ Q \ U \ V]^T
\]  \hspace{1cm} (4.5)

where \([\ ]^T\) denotes transpose.

Each element of the Stokes matrix can be written in terms of the light intensities as

\[
I = I_H + I_V
\]

\[
Q = I_H - I_V
\]

\[
U = I_P - I_M
\]

\[
V = I_R - I_L
\]  \hspace{1cm} (4.6)
where \( I_H, I_V, I_P, I_M, I_R, I_L \) are the light intensities measured with a horizontal linear polarizer, a vertical linear polarizer, a +45° linear polarizer, a -45° linear polarizer, a right circular analyzer and a left circular analyzer respectively placed in front of the detector.

The Stokes elements of a single simple wave are related by

\[
I^2 \geq Q^2 + U^2 + V^2
\]

The equality holds if the light is polarized. If the light is unpolarized, \( Q = U = V = 0 \). Natural sunlight is an example of completely unpolarized light and so it is represented as \([ I, 0, 0, 0 ]^T\). In general, however, light is partially polarized and so it consists of both polarized and unpolarized components. This leads to the concept of degree of polarization (DOP) given by \( \left( Q^2 + U^2 + V^2 \right)^{1/2} / I \), the degree of linear polarization (DOLP) defined as \( \left( Q^2 + U^2 \right)^{1/2} / I \) and the degree of circular polarization (DOCP) equal to \( V / I \).

The Stokes parameters for linearly polarized and circularly polarized light are shown in Table 4.1.
Table 4.1  Stokes Parameters for Polarized light

<table>
<thead>
<tr>
<th>Linearly Polarized</th>
<th>Circularly Polarized</th>
</tr>
</thead>
<tbody>
<tr>
<td>$0^\circ$</td>
<td>$90^\circ$</td>
</tr>
</tbody>
</table>
| \[
\begin{pmatrix}
1 \\
1 \\
0 \\
0
\end{pmatrix}
\] | \[
\begin{pmatrix}
1 \\
-1 \\
0 \\
0
\end{pmatrix}
\] | \[
\begin{pmatrix}
1 \\
0 \\
1 \\
0
\end{pmatrix}
\] | \[
\begin{pmatrix}
1 \\
0 \\
-1 \\
0
\end{pmatrix}
\] |
| Right               | Left                |
| \[
\begin{pmatrix}
1 \\
0 \\
0 \\
1
\end{pmatrix}
\] | \[
\begin{pmatrix}
1 \\
0 \\
0 \\
-1
\end{pmatrix}
\] |
4.2.7 **Mueller Matrices**

The state of polarization of a beam is changed on interaction with an optical element. It is possible to represent such optical elements by a $4 \times 4$ matrix called the Mueller matrix. The Mueller matrix describes the relation between "incident" and "transmitted" Stokes vectors and is written as

$$ S_{\text{out}} = M S_{\text{in}} \quad (4.7) $$

where $S_{\text{in}}$ and $S_{\text{out}}$ are the incident and the output Stokes vector respectively.

Mueller matrix formulation is very useful since it gives us a simple means of determining the state of polarization of a beam transmitted by an optical element for an arbitrarily polarized incident beam. Moreover, if a series of optical elements are interposed in a beam, the combined effect of all these elements may be determined by merely multiplying their associated Mueller matrices. The Mueller matrix can be obtained experimentally by measurements with different combination of source polarizers and detection analyzers.

4.2.8 **Absorption and Scattering by a sphere**

The most important exactly soluble problem in the theory of absorption and scattering by small particles is that for a sphere of arbitrary radius and
refractive index. Gustav Mie developed a formal solution to this problem, which is now called the Mie theory. Mie theory of scattering is a straightforward application of Maxwell's equations to an isotropic, homogenous, dielectric sphere. It is equally applicable to spheres of all sizes, refractive indices and for radiation at all wavelengths.

Mie's classical solution is described in terms of two parameters, $n_r$ and $\alpha$. $n_r$ is the magnitude of refractive index mismatch between particle and medium. This is expressed as the ratio of the refractive index for particle and medium as

$$n_r = \frac{n_p}{n_{med}} \quad (4.8)$$

The size of the surface of refractive index mismatch is expressed as size parameter $\alpha$. This is the ratio of the meridional circumference of the sphere ($2\pi a$, where $a$ is the radius) to the wavelength ($\lambda/n_{med}$) of light in the medium. It is given by

$$\alpha = 2\pi a / (\lambda / n_{med}) \quad (4.9)$$

Consider a source, a spherical scattering particle and an observer whose three positions define a plane called the scattering plane. Incident light and scattered light can be reduced to their components, which are parallel or perpendicular to the scattering plane. As shown in the Figure 4.3, the parallel
Figure 4.3  Scattering of polarized light
and perpendicular components can be experimentally selected by a linear polarizer oriented parallel or perpendicular to the scattering plane.

The Scattering matrix describes the relationship between incident and scattered electric field components perpendicular and parallel to the scattering plane as observed in the "far-field"

\[
\begin{bmatrix}
E_{is} \\
E_{ls}
\end{bmatrix} = \begin{bmatrix}
S_2 & S_3 \\
S_4 & S_1
\end{bmatrix} \begin{bmatrix}
E_{in}
\end{bmatrix}
\begin{bmatrix}
\exp(i k (r-z)) \\
-ikr
\end{bmatrix}
\]

(4.10)

The above expression simplifies in practical experiments:

- The exponential term, \(-\exp(ik(r-z))/ikr\), is a transport factor that depends on the distance between scatterer and observer. If one measures scattered light at a constant distance \(r\) from the scatterer, eg., as a function of angle or orientation of polarization, then the transport factor becomes a constant.

- The total field \((E_{\text{tot}})\) depends on the incident field \((E_i)\), the scattered field \((E_s)\), and the interaction of these fields \((E_{\text{int}})\). If one observes the scattering from a position which avoids \(E_i\), then both \(E_i\) and \(E_{\text{int}}\) are zero and only \(E_s\) is observed.
For "far-field" observation of $E_s$ at a distance $r$ from a particle of diameter $d$ such that $kr >> n_c^2$, $k = 2\pi/\lambda$, $n_c = d/\lambda$, the scattering elements $S_3$ and $S_4$ equal zero (Eq. 4.75, Bohren and Huffman).

Hence for practical scattering measurements, the above equation simplifies to the following:

$$\begin{bmatrix} E_{\nu} \\ E_{\lambda} \end{bmatrix} = \frac{\exp(ik(r - z))}{-ikr} \begin{bmatrix} S_2 & 0 \\ 0 & S_1 \end{bmatrix} \begin{bmatrix} E_{\nu} \\ E_{\lambda} \end{bmatrix}$$

(4.11)

Maxwell's equations are solved in spherical coordinates through separation of variables. The incident plane wave is expanded in Legendre polynomials, so the solutions inside and outside the sphere can be matched at the boundary. The far field solution is expressed in terms of two scattering functions, [142]

$$S_1 = \sum_{n=1}^{\infty} \frac{2n+1}{n(n+1)} \left[ a_n \pi_n(\cos \theta) + b_n \tau_n(\cos \theta) \right]$$

(4.12)

$$S_2 = \sum_{n=1}^{\infty} \frac{2n+1}{n(n+1)} \left[ b_n \pi_n(\cos \theta) + a_n \tau_n(\cos \theta) \right]$$

where $\theta$ is the scattering angle.
The functions $\pi_n$ and $\tau_n$ are given by

$$\pi_n(\cos \theta) = \frac{P_2^1(\cos \theta)}{\sin \theta}$$

(4.13)

$$\tau_n(\cos \theta) = \frac{d}{d\theta} P_n^1(\cos \theta).$$

where $P_n^1$ are associated Legendre Polynomials of the first kind.

and

$$a_n = \frac{\psi_n'(m\alpha)\psi_n(\alpha) - m\psi_n(m\alpha)\psi_n'(\alpha)}{\psi_n'(m\alpha)\xi_n(\alpha) - m\psi_n(m\alpha)\xi_n'(\alpha)}$$

(4.14)$$b_n = \frac{m\psi_n'(m\alpha)\psi_n(\alpha) - \psi_n(m\alpha)\psi_n'(\alpha)}{m\psi_n'(m\alpha)\xi_n(\alpha) - \psi_n(m\alpha)\xi_n'(\alpha)}$$

(4.15)

$\alpha = ka = 2\pi a / \lambda$ is the size parameter, $m$ is the index of refraction, $a$ is the particle radii and $\psi_n$ and $\xi_n$ are spherical Bessel functions.

The Mie scattering matrix is given by,

$$P(\Theta) = \begin{bmatrix}
P_{11}(\Theta) & P_{12}(\Theta) & 0 & 0 \\
P_{12}(\Theta) & P_{11}(\Theta) & 0 & 0 \\
0 & 0 & P_{33}(\Theta) & P_{34}(\Theta) \\
0 & 0 & -P_{34}(\Theta) & P_{33}(\Theta)
\end{bmatrix}$$

(4.16)
where the four independent Mie scattering matrix elements are,

\[ P_{11}(\Theta) = \frac{2\pi}{k^2 \sigma_z} [|S_1(\Theta)|^2 + |S_2(\Theta)|^2] \]  
(4.17)

\[ P_{12}(\Theta) = \frac{2\pi}{k^2 \sigma_z} [|S_2(\Theta)|^2 - |S_1(\Theta)|^2] \]  
(4.18)

\[ P_{23}(\Theta) = \frac{2\pi}{k^2 \sigma_z} [S_2(\Theta)S_1^*(\Theta) + S_1(\Theta)S_2^*(\Theta)] \]  
(4.19)

\[ P_{24}(\Theta) = \frac{2\pi}{k^2 \sigma_z} [S_2(\Theta)S_1^*(\Theta) - S_1(\Theta)S_2^*(\Theta)] \]  
(4.20)

The Mie extinction, scattering and absorption cross-sections can all be expressed in terms of the \( a_n \) and \( b_n \) coefficients as,

\[ \sigma_e = \frac{2\pi}{k^2} \sum_{n=1}^{\infty} (2n + 1) \text{Re}(a_n + b_n) \]  
(4.21)

\[ \sigma_s = \frac{2\pi}{k^2} \sum_{n=1}^{\infty} (2n + 1)(|a_n|^2 + |b_n|^2) \]  
(4.22)

\[ \sigma_a = \sigma_e - \sigma_s. \]  
(4.23)

Another useful quantity is the asymmetry factor, \( g \), which is the first moment of the phase function and is given by,

\[ g = \int_{4\pi} P_{11}(\Theta) \cos \Theta \, d\Omega. \]  
(4.24)
The asymmetry factor describes the shape of the phase function; $g>1$ indicates forward scattering, while $g<1$ indicates backscattering. It is a useful parameter for characterizing the phase function independent of scattering angle. The Mie asymmetry factor can be expressed as,

$$
g = 2 \sum_{n=1}^{\infty} \left[ \frac{n(n+2)}{n+1} \text{Re}(a_n a_{n+1}^* + b_n b_{n+1}^*) + \frac{2n+1}{n(n+1)} \text{Re}(a_n + b_n^*) \right].$$

(4.25)

4.3 POLARIZATION FILTERING TO ENHANCE VISIBILITY OF SUBSURFACE TISSUE IMAGES

Recently there has been a considerable interest in using the polarization state of the light as a discrimination criterion. The polarization discrimination technique is based on the assumption that weakly scattered light retains its initial polarization whereas highly scattered light does not. Consider a tissue illuminated by a linearly polarized source. As photons penetrate the tissue, they interact with various tissue structures and some of the injected photons emerge from the tissue in the backscattering direction. Light that reflects from the surface (known as spectral reflection) is polarized but the light that backscatters from somewhere below the surface of the tissue is depolarized. Therefore light viewed in the orthogonal state contains a large proportion of the light that interacted with the object than the copolarized
light. The use of this fact to discriminate between short and long path photons in a scattering medium has been well explored theoretically and practically. Studies clearly indicate that polarization discrimination [112 - 127] can be effective in extending the depth of visibility.

However even for this orthogonal state, as tissue thickness increases, the loss of contrast is still the light backscattered by the medium. So contrast of the images obtained could be enhanced if this back scattered contribution from the surrounding medium is removed. Hence if the copolarized image is also recorded and a suitable fraction is subtracted from the orthogonal polarized image, significant contrast could be achieved.

In this chapter, a Monte Carlo model for the proposed method is simulated. Model images are compared with the experimental images at 632nm for a scattering phantom.

4.4 MONTE CARLO SIMULATION

The essential characteristic of Monte Carlo code is to study the propagation of light in biological tissue. The basic idea is that photons are launched from an isotropic point source of unit power within an infinite homogeneous medium with no boundaries. The medium has optical properties of absorption, scattering and anisotropy. N photons are launched, each with a
"photon weight" initially set to 1. The photon takes steps between interactions with the tissue. The steps are based on the probability of photon movement before interaction by absorption and scattering. During each step as the photon propagates, the photon deposits a fraction of its weight into the local bin at its position. The photon is assumed to be dead if the weight of the photon falls below a fixed threshold. Now a new alive photon is launched.

Photons are launched as a beam from the origin that initially propagates parallel to the z-axis. The illumination is assumed to be linearly polarized. The biological tissue is modeled as a slab infinite in the x-y plane containing randomly distributed spherical Mie scatterers with size parameter \(ka\) where \(k\) is the wave number and \(a\) is the particle radius [118,119]. The scattering medium is assumed to be homogenous; hence the probability density for the distance traveled by the ray before it encounters a particle is determined by a negative exponential distribution with mean equal to the mean free path. Hence, in the simulation model the distance between two successive scattering points is a random number chosen from this distribution.

After every scattering event in the tissue, the polarization state and the direction of propagation of the ray changes. The Stokes vector carries the full polarization information of the wave under simulation. The scattering angle is chosen randomly according to a probability law depending on the phase function [117]. The scattering angle, \(\theta\) is generated by inversion and rejection.
methods [143]. At each scattering position, the code randomly chooses the azimuthal angle, $\psi$. Bruscaglioni et al [114] have proved that if the number of effective scattering events were larger than a few units and for particulates that were not small in comparison with the wavelength, the angle $\psi$ could be assumed to be equiprobable. Finally, performing the local coordinate transformation using the rotation matrix $T$ [117], the modified Stokes vector after scattering by the particle is

$$S_{H_{\text{out}}} = T(-\alpha) M(\theta) T(\psi) S_{H_1}$$  \hspace{1cm} (4.26)

where $M(\theta)$ is the Mie scattering matrix, and $\alpha$ is the rotation angle with respect to the ray direction [117]. Figure 4.4 illustrates the ray directions and angles $\alpha$, $\theta$, $\psi$ for a ray having initial direction $d$ and scattered to a new direction $d'$. $(\theta, \phi)$ and $(\theta', \phi')$ are the spherical polar coordinates of $d$ and $d'$.

To image an object located below the surface of the tissue, it is assumed that the object is placed at the boundary of the tissue i.e., at a distance $z_1$ in the $z$ direction. The object is modeled as a depolarizing diffuse reflector. The depolarization caused by the object can be modeled by generating random numbers for the Stokes parameter. But this is equivalent to inducing random rotation in $Q, U, V$ and is modeled by the Mueller matrix [144] given by,
Figure 4.4  Illustration of angles $\alpha$, $\theta$ and $\psi$
\[ M = \begin{pmatrix} 1 & 0_3^T \\ 0_3 & R(B,C,D) \end{pmatrix} \] (4.27)

where \( R(B,C,D) \) is a Euler matrix of dimension 3x3, \( 0_3 \) is a zero three column vector and \( B,C,D \) are Euler angles.

The photons after scattering by the tissue medium hit the object and exit the tissue. For each photon backscattered and returning to the top surface, the coordinates \((x_d, y_d)\) at which it intersected the detector plane were computed using geometrical ray tracing [62] using

\[
x_d = -x + \alpha_x (f - h) \tag{4.28}
\]
\[
y_d = -y + \alpha_y (f - h) \tag{4.29}
\]

where \( \alpha_x \) and \( \alpha_y \) are the exit angles onto the \(xz\) and \(yz\) plane respectively and can be computed from the directional cosines

\[
\alpha_x = \mu_x / |\mu_x| \tag{4.30}
\]
\[
\alpha_y = \mu_y / |\mu_z| \tag{4.31}
\]

and \( f \) and \( h \) are the focal length and distance of the lens of the imaging system from the surface of the tissue medium. The equation for \(x_d\) and \(y_d\) takes into account the refraction at the tissue-air interface.
A photon is counted as detected if its position falls within the area of the imaging lens, which is assumed to be circular and is determined by

\[(x_d^2 + y_d^2)^{1/2} < d_p/2 \quad (4.32)\]

where \(d_p\) is the lens diameter.

The radiance components parallel to the reference plane i.e., copolarized and perpendicular to the reference plane i.e., cross polarized components can be expressed in terms of the Stokes parameters as

\[E = (1 + Q)/2 \quad \text{and} \quad H = (1 - Q)/2 \quad (4.33)\]

The first two rows of the Stokes vector of these photons impinging on the lens yield the copolarized and the cross polarized components and these are transformed to the image plane using paraxial optics. The image plane was modeled as a grid with each pixel of size 0.04 x 0.04 cm and yields an image of size 60 x 60 pixels.

4.5 THE SIMULATION MODEL

A linearly polarized thin beam at 632 nm impinges on the scattering tissue medium. The scattering medium was considered to be a homogenous suspension of 2.02 μm diameter polystyrene spheres in water. The density of
scatterers is assumed to be 0.001 per cubic micron. The refractive index of the medium (water) was assumed to be 1.33. At 632nm, the complex index of refraction of the spheres was 1.589 \text{--} i0.00113 and the sample's scattering coefficient, reduced scattering coefficient, anisotropy and absorption coefficient were 100\text{cm}^{-1}, 8.9\text{cm}^{-1}, 0.911 and 1.33\text{cm}^{-1} respectively. These values have been chosen so as to set to values obtained from real muscle tissue [37].

The optical detector is a CCD camera with a circular lens of radius 0.05\text{cm} placed at a distance of 0.1\text{cm} from the surface of the tissue medium. The focal length is also assumed to be 0.1\text{cm}.

### 4.6 EXPERIMENTAL SET UP

A schematic of the experimental apparatus used for obtaining the images is shown in Figure 4.5. A He Ne laser (Uniphase Inc, 1101P) emitting approximately 4mW of 632nm light was first linearly polarized and then focused onto the tissue phantom. The phantom that was utilized in the study was a suspension of 2.02 \text{\textmu m} diameter polystyrene spheres (Polysciences Inc. 19814). The sample was
Figure 4.5  Schematic of the proposed method

LS – He Ne laser  
CCD - Imaging camera  
LP - linear polarizer
created by dilution with distilled water to 0.001 spheres per cubic microns. The index of refraction of the spheres was $1.589 - 0.00113$ and the refractive index of the medium (water) is 1.33. The sample had a scattering coefficient ($\mu_s$) of 100 cm$^{-1}$, reduced scattering coefficient ($\mu'_s$) of 8.9 cm$^{-1}$, anisotropy ($g$) of 0.911 and absorption coefficient ($\mu_a$) of 1.33 cm$^{-1}$ at 632nm, equal to the values used in the simulation.

The sample was held in a glass container with flat walls. A 4mm diameter and 1mm thick object composed of chalk material was positioned at the bottom of the sample. Light emerging from the sample and object was collected though a camera lens and then into the second polarizing element. The image was then focused onto a digital CCD camera. The thickness of the phantom sample was varied from 0.5cm to 2 cm and for each thickness two images, namely cross polarized image and copolarized image were acquired. The obtained images were processed.

4.7 RESULTS AND DISCUSSION

4.7.1 Simulation results

The simulation for the imaging system illustrated in Figure 4.5 was run by launching a beam of photons from the origin. Random values were taken for the scattering distance, scattering and the azimuth angles as described in section 4.4. The depth of the object was incremented in steps of 0.5 cm, and
Figure 4.6 Images of object from simulation for object depths of 1 and 1.5cm: (a) orthogonal polarized image; (b) co-polarized image; (c) image after polarization filtering
copolarized and cross polarized images were recorded at 632nm. The simulation was run for a launch of $1 \times 10^7$ photons.

In Figure 4.6, (a) and (b) shows the orthogonal and copolarized images of the object at 632 nm from the simulation. The images in the first and second row are for the object at depth of 1cm and 1.5cm respectively. Figure 4.6 (c) is the image after subtraction of orthogonal state and fraction of copolarized state. In performing the subtraction [145,146], if the subtraction resulted in a negative value, it was made equal to zero or a black pixel. Clearly the improvement in contrast of the images in 4.6(c) over 4.6(a) and (b) is visible proving the merit of this model. The optimum subtraction fraction was chosen as 0.5 based on best contrast of the final image.

4.7.2 Measurement of Contrast

To validate the proposed technique, image contrast as a function of depth for the images obtained was computed. The contrast is defined as

$$\text{Contrast} = \frac{(A_1 - A_B)}{(A_1 + A_B)}$$

(4.34)

where $A_1$ is the average intensity from the light returning from the object calculated from the central 3 x 3 pixels and $A_B$ is that from light returning from the background calculated from the central 21 x 21 pixels. The image
contrasts are listed in Table 4.2 for three different depths. The values in last column show the improvement of contrast by the proposed method.

**Table 4.2** Contrast for the simulated images of the object obtained at depths of 0.5, 1.0 and 1.5 cm in the tissue.

<table>
<thead>
<tr>
<th>Depth of object</th>
<th>Cross polarized image at 632nm</th>
<th>Proposed method</th>
</tr>
</thead>
<tbody>
<tr>
<td>0.5</td>
<td>0.901</td>
<td>1.0000</td>
</tr>
<tr>
<td>1.0</td>
<td>0.6293</td>
<td>0.7287</td>
</tr>
<tr>
<td>1.5</td>
<td>0.5300</td>
<td>0.6511</td>
</tr>
</tbody>
</table>

### 4.7.3 Experimental Results

The experiment was carried out using the experimental set up and cross and copolarized images of the object at varying depths were recorded. Figure 4.7(a) and 4.8(a) shows the orthogonal image from the experimental set up for the object at a depth of 0.5 and 1cm. Figure 4.7(b) and 4.8(b) is the image after subtraction of orthogonal state and fraction of copolarized state for the respective depths.
Figure 4.7  Images of the object from experimentation at a depth of 0.5cm: (a) cross polarized image; (b) image after polarization filtering
Figure 4.8 Images of the object from experimentation at a depth of 1 cm: (a) cross polarized image; (b) image after polarization filtering
4.7.4 Intensity Profile

The intensity profile of the experimental images along the line containing the object for varying depths is shown in Figure 4.9. It can be seen that the profile is narrow for the object when it is at a depth of 1cm from the surface and the profile widens as depth is increased to 1.5cm. When the depth of object is further increased, the object is no longer visible even by the proposed method.

![Intensity profile of experimentally obtained images after polarization filtering at varying depths](image-url)

Figure 4.9 Intensity profile of experimentally obtained images after polarization filtering at varying depths
4.8 SUMMARY

In this chapter the first phase in the development of an optical imaging system is discussed. This phase termed polarization filtering enhances the contrast of tissue images at deeper depths using image-processing tools. Image subtraction is done to detect changes in the tissue composition. Computer simulations based on Monte Carlo model have been carried out and experimentation done to check the validity of the method. Resultant images indicate the merit of the proposed method for depths up to 1 cm.