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Research Paper

PERFORMANCE OF DIFFUSE OPTICAL TOMOGRAPHY USING DUAL SOURCE ILLUMINATION: RECONSTRUCTION OF OPTICAL PARAMETER FROM REAL PARTS OF THE MEASURED DATA

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We study the performance of a diffuse optical tomography system using phase-correlated illumination, by analyzing the contribution from the real and imaginary parts of the data. After introducing the iterative reconstruction algorithm, which implements the Jacobian of the dual source system for data types real and the complex, we present results of reconstruction from such data types. It is shown through simulations, that the real part of the data, and to a lesser extent the magnitude leads to noise cancellation and improved localization and resolution of inhomogenity in the reconstruction. However, the imaginary part of the measurement data, increases the noise content of the reconstruction. The minimum size of the inhomogenity detect from phase-correlated data (real part) is 4 mm in diameter.

Keywords: Diffuse optical tomography, Phased array, Jacobian, MoBIIR algorithm

INTRODUCTION

Diffuse Optical Tomography (DOT) is a process of reconstructing the internal structure of a tissue using measurements obtained by illuminating a tissue with Near-Infra Red (NIR) light. The light transport in the tissue is characterized by two optical parameters, namely the absorption and scattering coefficients. Using DOT we can detect cancerous part in breast. If you have cancer in breast, then that particular position which will absorb more light and it will change the absorption and scattering coefficient. Main advantage of DOT, radiation is noninvasive and non-ionizing so we can use repeatedly and it is inexpensive and portable also.

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In DOT imaging one illuminates the tissue with an array of light sources and measures the light leaving the tissue with an array of detectors. For each source location, one records a quantity proportional to the light reaching each detector. A model of the propagation of light in tissue is developed and parameterized in terms of the unknown scattering and/or absorption as a function of position in the tissue. Using the model, with measurement data on these detectors, for each of the source positions, one attempts to "invert" the propagation model to recover the parameters of interest, or, in other words, to estimate the scattering and/or absorption parameters.

The combination of optical properties has two important implications (1) low absorption allows the transmission of NIR light in tissue to about 8 cm. (2) High scattering coefficient reduces the contrast and resolution of images formed with light.

In this paper, we are concerned with the DOT with a frequency modulated light source, which is called the frequency domain DOT. Here the sinusoidally modulated light source is used to launch a photon density wave. If the medium is homogenous, photon density wave travels at constant velocity and is exponentially attenuated as it propagates. The attenuation values and phase delay of the wave are functions of absorption and scattering coefficients of the medium. For inhomogenous tissue, the presence of spectroscopic variations in the absorption and scattering coefficients results in the deformation of photon density wave. The distinct advantage of frequency domain method is that the value of the amplitude and phase can be easily measured and the location and properties of the inhomogenities in tissue can be easily reconstructed to identify the portion affected by pathology.

One of the methods in DOT, which uses the amplitude cancellation technique, is the phased array system. This method uses dual out of phase sources (phased array), which can detect and locate small inhomogenities in turbid medium. This technique provides a better noise tolerance and contrast enhancement. The principle of operation of dual source illumination DOT is that object is illuminated by two intensity modulated light beams, which are out of phase. The interference of these two sources creates a null plane at the mid-point between the sources and a phase shift in a homogeneous object. The presence of inhomogeneous inclusion shifts the position of the null-plane and the phase; these data lead to sensitive detection of inhomogenities. The analytical solution of dual source system can be obtained by applying the summation theory to the single source system because of the linearity of the problem and can be extended to tomographic image reconstruction.

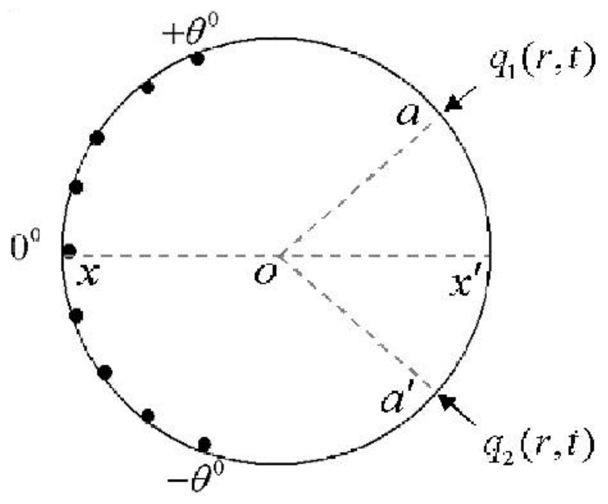
DUAL SOURCE ILLUMINATION

The dual-source 'phased-array' system considered here has provision to illuminate the object with two frequency modulated light sources that are phase-correlated with phase shift of between them. They are given by:

$$q_1(r, t) = A_{dc} + A_{ac}\cos(\mu_0 t)$$
$$q_2(r, t) = A_{dc} - A_{ac}\cos(\mu_0 t) \quad ...(1)$$

The geometry of data collection is shown in Figure 1, wherein we consider a 2-d object

Figure 1: Dual Source Out-of-Phase Illumination of the Tissue with a Detector Located on Opposite Side of the Object Phantom Along the Line Bisecting the Angle of Separation of the Sources



(the cross section of cylinder), which has number of detectors, placed diametrically opposite side with respect to the source (Figure 1). These signals are modulated by high frequency (100 MHz) AC signal over the DC bias. The reason for choosing this out of phase dual interfering illumination is that it produces highest sensitivity in detection of inhomogenity.

The solution to the photon transport problem is obtained by solving the Diffusion Equation (DE) given below.

$$\left[-\nabla \cdot k(r)\nabla + \left(«_a(r) + \frac{j\mu /}{c}\right)\right] ¤(r,\, \mu) = \tilde{Q}(r,\, \mu)$$

The detected output is of the form A_{out-dc} + A_{out-ac} cos($\mu_0 t$ + ¤), from which we can optain the dc response, ac response and phase response. For the dual-source phased array illumination with detectors located to span the opposite half boundary on the detector system as shown in Figure 1, we obtain amplitude and phase response at these detectors. Figure 2 gives amplitude and phase response for modulation frequency of 100 MHz for a homogenous object. From the magnitude response it can be seen there is a 35 dB drop in intensity, and 1800 phase shift along the null plane. Similarly for a inhomogeneous object the drop in intensity is less than 10 db and shifts the null point (Figure 3). This plane where we get a 35 dB drop in intensity for a homogenous object is known as the cancellation plane, and the amplitude is minimum on this plane and the phase response undergoes a phase transition along this plane.

The presence of an inhomogeneity in optical properties shifts this null point, and forms the basis for highly sensitive phased-array DOT imaging system.

Figure 2: Amplitude and Phase Response for Homogenous Object with a Dual Source Illumination System

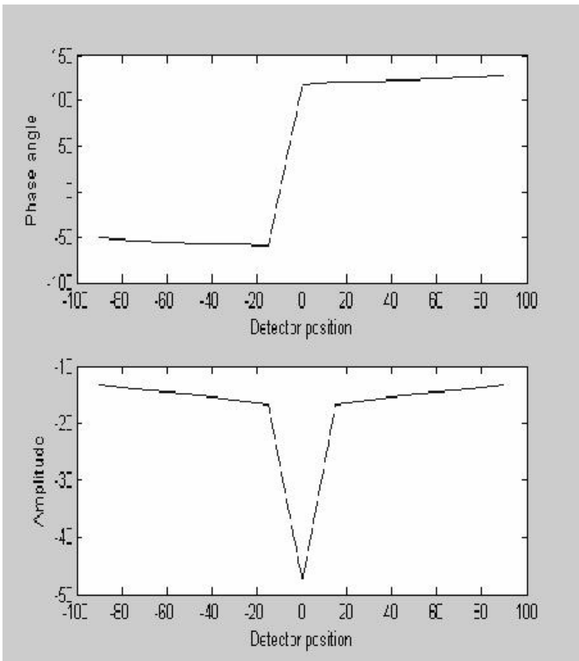
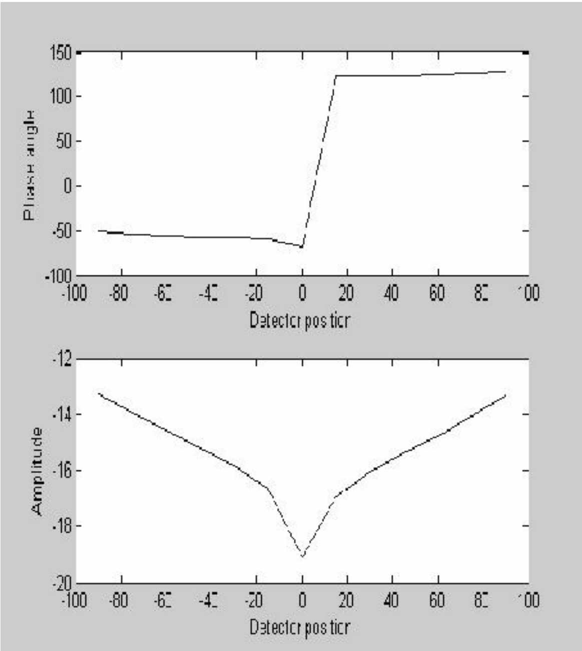


Figure 3: Amplitude and Phase Response for Inhomogeneous Object with a Dual Source Illumination System



In simulations, we considered two constant intensity sources given by,

$$q_1(r, t) = A_0$$

$$q_2(r, t) = -A_0 \quad ...(2)$$

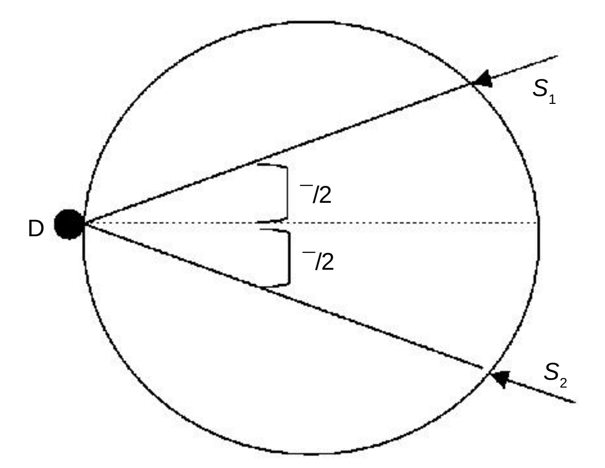
illuminating the tissue phantom. It was found that the response of the system to a dual source of equal intensity but of differing sign as given in Equation (2) was comparable to a system with a frequency-modulated source as given in Equation (1). Hence it was decided to use a constant intensity dual source with a phase difference ® radians in place of the phase correlated frequency modulated source for simulations.

DATA GATHERING TECHNIQUE

The dual source of constant intensity with 100 MHz and separated by ® radians is used in place of the phase correlated frequency modulated source. A detector is placed on the null-plane. As referred before due to the out of phase nature of the input signals the received signal at the detector is zero for a homogenous object. In addition, the common mode noise picked up by detector also gets cancelled. This feature enhances the noise tolerance of the phased-array imaging system. The presence of an inhomogenity unsettles this null in the mid-plane and the detector reads amplitude and phase shift. Phase shift is related to the extra path length of the ray inside the inhomogenity.

The data gathering technique for the dual source is shown in Figure 4. The two sources, S_1 and S_2 are separated by ¯ ⁰ and the detector is placed on the opposite side of the phantom on a line bisecting. Data can be gathered by rotating the source-detector in unison in equal angle as steps so that it covers the entire phantom. The other method is to keep the source and detector in a fixed position and rotate the phantom and obtain the measurement readings. Normally the first

Figure 4: Data Gathering Geometry for the Dual Source Phased-Array System

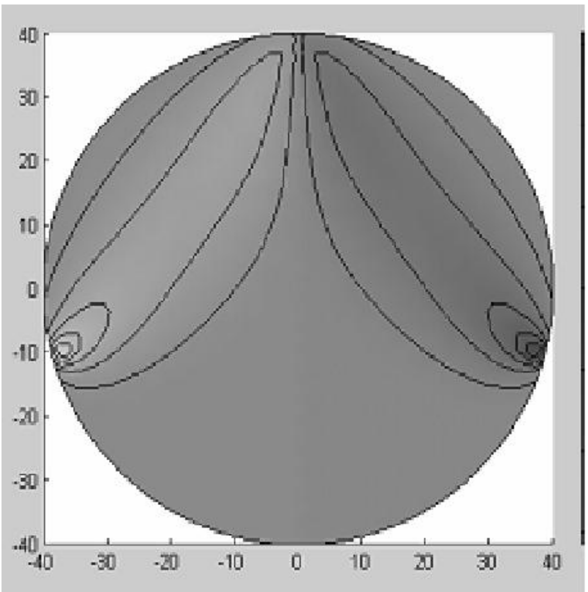


method is preferred which replicates the geometry used in similar clinical systems.

RECONSTRUCTION PROCEDURE

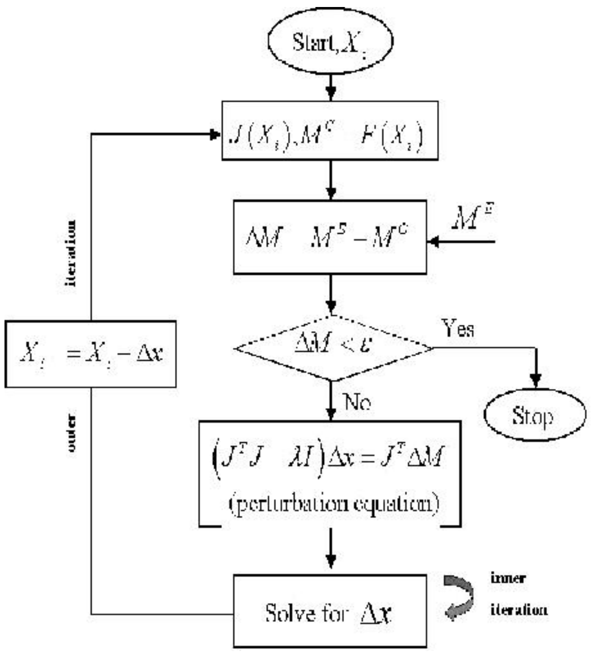
An image of tissue optical properties has been reconstructed using data points acquired using a source with wavelength 785 nm and by finite element based calculation of diffusion equation. For reconstruction we use the iterative linear MoBIIR algorithm. The Jacobian for the dual source system is obtained by algebraic addition of the Jacobians for the two sources computed separately. Typical Jacobian for dual sources S_1 and S_2 separated by $^{-\,0}$ for absorption coefficient ma is given in Figure 5.

Figure 5: The Jacobian for Two Sources of Strength S_1 = +1 and, S_2 = –1 with a Single Detector D



The MoBIIR algorithm used for reconstruction is shown in Figure 6. This algorithm can accept data measured on the boundary and reconstruct the optical properties of the object phantom, using inner and outer iterations in the flowchart (Figure 6).

Figure 6: MoBIIR Algorithm



COMPUTER SIMULATIONS

As considered earlier the object considered for our simulation study is a 2-d cross-section of a cylindrical object of diameter 80 mm with background absorption $\mu^b{}_a$ and reduced scattering coefficients $\mu^b{}_s$ of 0.02 mm^{-1} and 0.5 mm^{-1} respectively. Cylindrical absorption coefficient inhomogenities (i.e., scattering coefficient of the background is retained) of different diameter are introduced at different positions to create an inhomogeneous object for use in the numerical experiments. PDE toolbox available in Matlab is used for simulations. The circular domain representing phantom is discretized into 2880 non-overlapping triangular elements connected by 1501 nodes, and the forward model solves the DE for photon flux everywhere including the detector.

Data collection technique for the dual-source system is as shown in Figure 4. The two sources are separated by an angle ¯, which is increased in steps of 30° starting with ¯ = 30° to ¯ = 180°. As a result six measurements are obtained for a single detector position. The sources-detector combination in each of these angular separations is rotated around the phantom in unison in steps of 30°, covering 360° in 12 steps. Thus we obtain 72 detector readings totally.

RESULTS AND DISCUSSION

The phantom used for our study is shown in Figure 7a. For this object we have gathered the intensity data at the null point at 72 detector locations. This set of data was used to reconstruct the absorption coefficient distribution of the object using the algorithm shown in Figure 6. The data set was split into two sets: (i) Only the Real data, (ii) Real and Imaginary together to form complex set of data as measured. These are reconstructed using Jacobian estimated from these data types. The reconstruction using Real part of the data is shown in Figures 7b and 7c shows the same using the complex data. The results show that reconstruction has better noise tolerance and is capable of localising the position of the inhomogenity quite well. Comparing the results of Figures 7b and 7c one sees that the contribution from the imaginary part does not improve the quality of reconstruction. It was also found that the size of the inhomogenity was slightly enlarged due to the diffusive nature of the DOT, which exhibits low pass characteristics, which widens the inhomogenity. This phased array system could reconstruct as small as 4 mm inhomogeneity,

Figure 7a: Cross Section of Phantom with Inhomogenity Size of 4 mm and the Inhomogenity in Phantom has $_a$ of 0.04 and Background $_a$ of 0.02

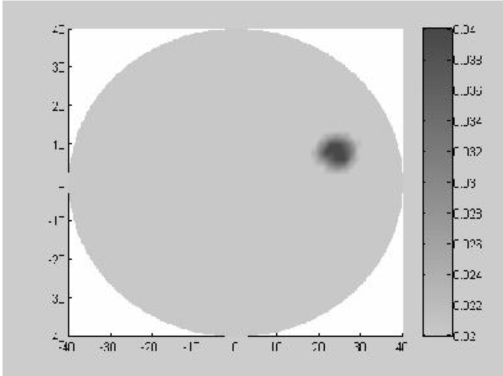


Figure 7b: Cross Section of Reconstructed Image of Phantom with Inhomogenity Using Real Part of Jacobian

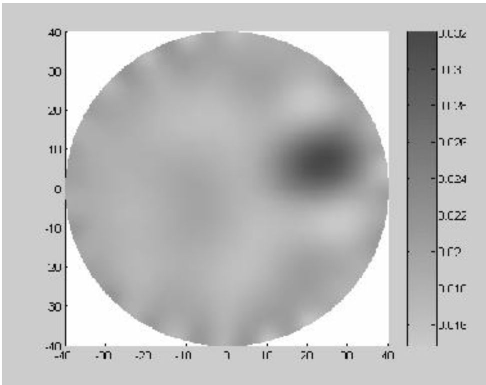
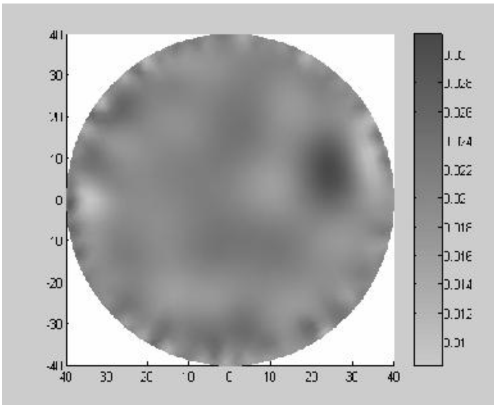


Figure 7c: Iimage of Phantom with Inhomogenity Using Real and Image Part



which is on par with the size of tumor of benign nature.

CONCLUSION

The result of reconstruction for a dual source DOT system is discussed. We have been able to properly reconstruct inhomogenities of diameter as small as 4 mm. This is usually not possible from a single source DOT system. It is shown that the noise cancellation is optimal with real part of the data and also plays a major role in localising the position of inhomogenity. The imaginary part is found to add the noise.

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