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Analysis of 3Dimensional Reconstruction in a MR-guided NIR Tomography System

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Astract: The algorithms for near-infrared (NIR) diffuse optical tomography within magnetic resonance imaging (MRI) of the female breast are analyzed in both circular and curved slab geometries.

OCIS codes: (170.3830) Mammography; (170.6960) Tomography; (100.3010) Image reconstruction techniques

1. Introduction

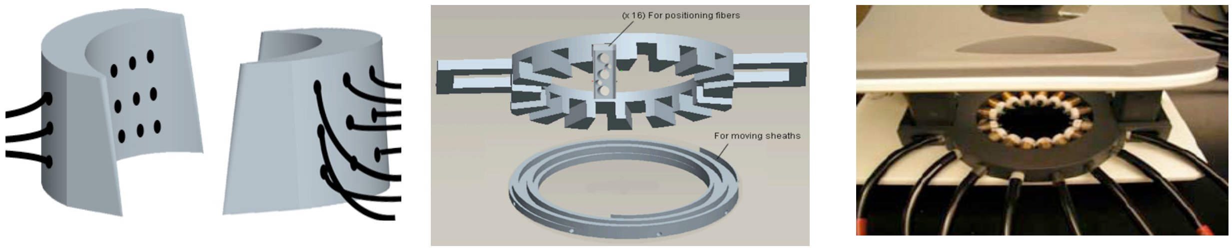
Near Infrared (NIR) optical tomography is used mainly to image brain and breast using NIR light as the signal[1, 2]. Typically, NIR light is delivered and collected on the boundary of the tissue volume. Using these boundary measurements, recovery of spatial distribution of optical properties (μ_a and $\mu_s^{/}$) is done using a non-linear iterative procedure. These optical parameters can be used in determining the tissue hemoglobin and water contents[2]. NIR imaging has the advantage of getting these functional parameters as it is a non-invasive and non-ionizing mechanism. Due to diffusive nature of the light, the spatial resolution of NIR imaging (~ 4-6mm) is quite poor. MRI has high spatial resolution (~ 1 mm), but lacks further ability to provide functional information. To take advantage of both imaging modalities, a dual imaging modality, combining both NIR with MRI has been developed at Dartmouth[3] and is being used in an ongoing clinical trial for breast cancer imaging. In this work, a technique which uses the spatial *a priori* structural information available from MRI for co-registering with a NIR image reconstruction procedure is quantitatively assessed. Note that in this type of regularization, we penalize the imaging problem with possible variations within the regions of same property (termed as *soft-priors*). The exact formulation of this regularization matrix is a Laplacian-structure described by Brooksby et. al[4], allowing intra-region variability to exist at a low level, and allowing larger discontinuity changes between property regions. This approach can work well even if the confidence in structural *a priori* information is low, as it is not applied in a highly forced manner, but rather coerces the pre-defined regions to have relatively smooth property values in the image.

However given this approach to regularization, it is imperative to analyze the capability to image heterogeneities in the imaging field, which may not be provided by the MR image. Also, the ability to use this approach in 3 dimensions is important as the MR provides fully volumetric images, which can then be used for the full 3 dimensional mesh grid. In this study, the 3D reconstruction issues are examined.

2. Methods

2.1 System Design

The updated MR-NIR system is based off a previous design implemented by Brooksby et al [1], with a newly added patient interface. A patient lies inside a 3.0T MRI (Philips Medical) with breast pendent through the patient interface, housing optical fibers for NIR data collection. MRI and NIR data are acquired simultaneously and used in the image reconstruction. Six wavelengths from 660-850nm are amplitude modulated at 100 MHz to provide amplitude and phase information. The patient interface being examined would utilize a curved-slab geometry which will provide for greater signal than a circular geometry, and more patient comfort than a traditional flat slab geometry. Here 48 optical fibers are planned to be positioned across the two surfaces in three layers and images are reconstructed in 3D. Work is being done to incorporate this design into a custom made MRI breast coil.



(a) (b) ©

Figure 1. (a,b) Two geometries considered for patient interface of the MR-NIR. (a) circular slab geometry (b) circular ring geometry shown with 16 sheaths used to position 3 fibers each and a spiral used to move all sheaths concentrically (c) Current patient interface with ring geometry in MRI breast coil

2.2a Image Formation

A frequency domain Diffusion Equation can be used to model NIR transport through tissue at reasonably large distances, as long as scattering dominates over absorption. The Diffusion Equation is used to simulate measured optical signals based on specified absorption and scattering properties. It is given as

$$-\nabla \cdot D(r)\nabla\Phi(r,w) + \left[\mu_a(r) + \frac{iw}{c}\right]\Phi(r,w) = S(r,w) \qquad (1)$$

where $\Phi(r,\omega)$ is the photon density at position r and modulation frequency w. The diffusion coefficient is μ_t = $1/[3(\mu_a + \mu_s^{/})]$ and the isotropic source term is S(r,w). The reduced scattering coefficient is $\mu_s' = (1-g)\mu_s$. A type III boundary condition is used. Data measured from the system is solved with Finite Element Method (FEM) based reconstruction algorithm to generate the distributions of μ_a and μ_s'.

2.2b Incorporating priori information to image reconstruction

MRI structural data is incorporated along with the Tikhonov approach to Newton-minimization [5] to find the solution to:

$$(\hat{\mu}_a,\hat{D}) = \min_{\hat{\mu}_a,\hat{D}}\left\{\| y^* - F(\hat{\mu}_a,\hat{D})\| + \alpha \| L(\hat{\mu}_a,\hat{D}) - (\hat{\mu}_{a0},\hat{D}_0)\|\right\} \qquad (2)$$

The vector represented by 'y' is the experimentally measured data, and ||.|| represents the square root of the sum of the squared elements. The data model mismatch is handled by a regularization factor α, which is chosen empirically to be 0.1. Data from the MRI, either T1 or T2 weighted images, is labeled by like regions in terms of adipose and glandular tissue. The weighting matrix L is a filter matrix which essentially smoothes data in a particular region. The update equation in a reformulation from previous works [2] is:

$$\Delta\mu = \left(L^T L\right)^{-1} J^T \left[J\left(L^T L\right)J^T + \alpha I\right]^{-1}\Delta b \qquad (3)$$

where Δb is the difference between calculated and measured data. This formula results in a matrix that is more efficient in computational memory.

2.3 Phantom Studies

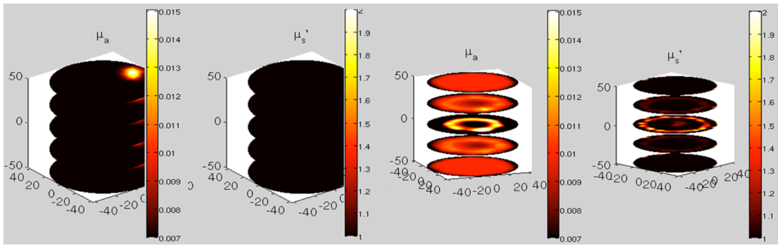
Real data was collected by imaging gelatin phantoms mixed with titanium oxide (*Sigma Inc.*) for scatter and India ink for absorption. Layers were made that included typical optical properties found for each type of breast tissue. Cylindrical phantoms had inclusions with higher absorption and scattering properties to simulate tumor properties. These phantoms were 86 mm in diameter and 100 mm in height.

3. Results

To show the improved accuracy of 3D reconstructions with prior structural information, simulated and phantom data sets were taken. In each of the reconstructed images, prior information yielded images with less noise. The MR-NIR reconstructed images also showed the ability to reconstruct features outside the plane of data. For both data sets, the plane of data was taken in the center plane of the cylinder.

3.1 Simulated Cylindrical Inclusion

A dataset was simulated on a cylindrical inclusion of diameter 15 mm extending the entire z axis, surrounded by a homogeneous background. The inclusion had double the contrast of the background for absorption. The inclusion was homogeneous for its scattering property. The optical properties for μ_a and μ_s' for the background were {0.01,1} respectively. For the inclusion, they were {0.02,1} for μ_a and μ_s'. As seen below, the prior reconstruction showed not only noise reduction from the non-prior image, but also information reconstructed off plane from the plane of data.

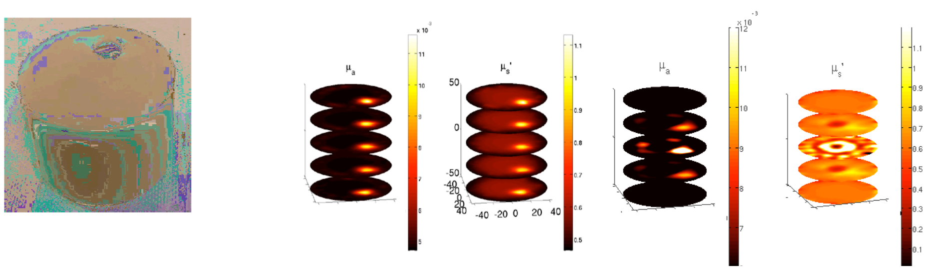


(a) (b)

Figure 2. Optical properties for 3D reconstructed data. (a) volume reconstructed with 3D prior information. (b) volume reconstructed without prior information.

3.2 Phantom Cylindrical 3 Layer

This dataset was recorded on our standalone NIR system which is currently in clinical trials [3]. The structural information was gathered from photographs taken of the phantom. An important consideration is the fact that although the photograph was accurate, it was very difficult to record the boundaries of the FEM mesh exactly. Even still, because the reconstruction technique does not impose exact boundaries on the reconstruction, we were able to properly obtain improved optical properties. This reconstruction example is more realistic in that there is a tumor inclusion within a glandular layer, which are surrounded by adipose tissue. The optical properties for μ_a and μ_s' for the outer fatty layer were {0.06,0.65} respectively. In the middle adipose layer, μ_a and μ_s' were {0.01,1}. The inclusion had μ_a and μ_s' of {0.02,1.2}.



(a) (b) (c)

Figure 3. Optical properties for the 3D reconstructed 3 layer phantom experiment. (a) Gelatin phantom imaged (b) volume reconstructed with prior information (c) volume reconstructed without prior information.

Here, the prior reconstruction technique shows more accurate results for both scattering and absorption. Also, noise is reduced considerably compared to not using prior structural information.

4. Discussion

Both the simulated and experimental data agree there is a need for prior structural information in order to successfully recover the optical properties with NIR optical tomography. Datasets reconstructed in 3D inherently have more noise and variation in the data than 2D datasets, and therefore require reconstruction techniques to control the noise. In comparing the simulated and experimental images, prior reconstruction techniques not only yield more consistent images, but it show that as more noise is added to the data with real experimental data, images cannot be resolved even on the plane of data without structural priors.

6. Acknowledgements

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7. References

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