Microwave Imaging for Breast Cancer Detection and Therapy Monitoring

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Abstract — Microwave imaging is based on recovering the electrical properties, namely permittivity and conductivity, of materials. Microwave imaging for biomedical applications is particularly interesting, because the available range of dielectric properties of different tissues can provide substantial functional information about their health. Breast cancer detection and treatment response monitoring are areas where microwave imaging is becoming a promising alternative/complementary technique to current imaging modalities, mainly due to the significant dielectric property contrast between normal and malignant breast tissues. In this paper, we present our latest clinical microwave imaging system along with some 2D and 3D reconstructed images from different phantom experiments and patient data.

Index Terms — Breast Cancer, Microwave Imaging, Therapy Monitoring, Two-dimensional and Three-dimensional Reconstruction Algorithms.

I. INTRODUCTION

Breast cancer is a serious health problem in the U.S. and according to American Cancer Society (ACS), over 40,000 women in the U.S. were expected to die in 2009 from breast cancer [1]. Thus, breast cancer has the second highest mortality rate of cancers after lung cancer among women in the U.S, and is also the most commonly diagnosed one besides skin cancer [1]. Statistics show that from 1990 to 2006, breast cancer death rates decreased by 3.2% per year among women younger than 50 [1]. This decline has been attributed to improvements in treatment, treatment response monitoring, and also early detection of breast cancer. Today, X-ray mammography is the standard clinical technique to detect breast abnormalities; nonetheless, the increased level of fibroglandular tissue in higher density breasts may obscure the presence of tumors and as a result, the overall sensitivity of mammography can be severely reduced. In addition, from patients’ and health care providers’ perspective, high false-positive rate of X-ray mammography can potentially lead to unnecessary and expensive surgical interventions. There are other imaging techniques for screening and detection of breast cancer in the clinic, such as magnetic resonance imaging (MRI), X-ray CT, and ultrasound (US). While all these tools may be more sensitive than mammography in selected populations, they suffer from low specificity and therefore cannot provide functional information about the molecular-level changes in the tissue.

In order to overcome these shortcomings, new imaging modalities should be developed as replacement and/or incremental techniques to improve both sensitivity and specificity of the current imaging systems. Microwave imaging has the potential to be such an imaging modality. More specifically, microwave imaging spectroscopy for biomedical applications reconstructs the electrical properties of tissues (permittivity and conductivity) over the microwave frequency range. Previous studies have shown that there is a significant dielectric property contrast between normal and malignant breast tissue [2-4]. Therefore, microwave imaging can provide substantial functional information about the breast tissue health and can also be used as a detection and treatment response monitoring tool for breast cancer.

In this paper, we present our microwave imaging system at Dartmouth Hitchcock Medical Center (DHMC) in Section II. Section III includes a brief description of the image reconstruction algorithm, followed by some reconstructed 2D and 3D images from various phantom experiments and patient data in Section IV.

II. MICROWAVE IMAGING SYSTEM

Figure 1 shows our current clinical microwave imaging system (MIST) located at the Advance Imaging Center at the Dartmouth Hitchcock Medical Center (DHMC) in Lebanon, NH. As illustrated in Figure 2, a patient lies prone on the bed and the breast is pendant in the imaging tank which is filled with a coupling medium comprised of a mixture of glycerin and water. The coupling liquid closely mimics the average constitutive parameters of the breast while minimizing the microwave scattering and enhancing coupling of electromagnetic energy [5]. The data acquisition system consists of 16 monopole antennas which can operate over the frequency range from 500 MHz to 3.0 GHz. The antennas are arranged in a circular form and are mounted on two separate plates which can move up and down independent of each other. This configuration enables us to collect both in-plane and cross-plane data for three dimensional data acquisition. Each antenna operates as both transmitter and receiver: while one antenna channel transmits the signal, the other 15 receive it and this process is repeated for each antenna, resulting in a total number of 240 measurements of the scattered field at each plane. For 2D imaging, only in-
plane data is acquired. That is, all 16 antennas have the same height and move in the same plane as shown in Figure 2. More specific details about our imaging system can be found in Meaney et al [6].

The second problem, known as the inverse problem, involves estimating the dielectric properties from known microwave excitations and measured data.

The dielectric properties of a material are embodied in the complex wave number squared, \( k^2 \):
\[
k^2(r) = \omega^2 \mu_0 \varepsilon(r) - j \omega \mu_0 \sigma(r)
\]
where \( r \) is the position vector in the imaging domain, \( \omega \) is the angular frequency, \( \mu_0 \) is the free-space magnetic permeability, \( \varepsilon \) and \( \sigma \) are the electrical permittivity and conductivity, and \( j = \sqrt{-1} \). The core of our image reconstruction algorithm is the inverse problem for which an iterative Gauss-Newton scheme is used to minimize the data-model misfit. That is the difference between the measured (acquired data) and computed electric field calculated using the forward model. The objective function \( \Omega \) can be written as
\[
\Omega = \left[ \Gamma^m - \Gamma^c(k^2) \right|^2_{\eta} + \left[ \Phi^m - \Phi^c(k^2) \right|^2_{\eta} + \lambda \left\| L(k^2 - k^2_0) \right\|^2_{\eta}
\]
where \( \Gamma^m \) and \( \Gamma^c \) are the phases of the measured and computed field values, respectively [7, 8]. Due to the non-linear and ill-posed nature of this problem, some additional constraints known as regularizations are also required. In equation 2, \( \lambda \) is the Tikhonov regularization parameter, and \( L \) is a positive definite, dimensionless regularization matrix. \( k^2_0 \) is a prior estimate of \( k^2 \) and \( \| \cdot \|_2 \) is the vector two-norm.

Given the Jacobian matrix \( J \), the objective function \( \Omega \) can be minimized iteratively with respect to the property update, \( \Delta k^2_\eta \), leading to
\[
J^T J + \lambda L^T L \Delta k^2_\eta = J^T \left[ \Gamma^m - \Gamma^c(k^2_\eta) \right] + \Phi^m - \Phi^c(k^2_\eta) - L^T L(k^2_\eta - k^2_0)
\]
where \( k^2_\eta \) is the vector \( k^2 \) at iteration \( \eta \) and is updated as
\[
k^2_{\eta+1} = k^2_{\eta} + \Delta k^2_{\eta}
\]

For simplicity, in our reconstruction algorithm the regularization matrix \( L \) in equation 3 is set to be an identity matrix, which applies the same weight to all nodes of the reconstruction mesh. Moreover, the prior estimate of the complex wave number squared is set to its value at the previous iteration, \( k^2_\eta \). This leads to equation 5 which is a simplified version of the update equation in (3):
\[
J^T J + \lambda I \Delta k^2_\eta = J^T \left[ \Gamma^m - \Gamma^c(k^2_\eta) \right] + \Phi^m - \Phi^c(k^2_\eta)
\]
In the reconstruction algorithm, the dual-mesh approach is used. That is, the forward problem is solved on a uniform rectangular (for 2D) or rectangular cuboid (for 3D) FDTD grid, whereas the dielectric properties are
reconstructed on a triangular (for 2D) or tetrahedral (for 3D) element mesh centered within the antenna array (Figure 4).

![2D and 3D FDTD grid and reconstruction meshes](image)

**Fig. 4.** 2D and 3D FDTD grid and reconstruction meshes: a) 2D rectangular FDTD grid, b) 3D rectangular cuboid FDTD grid, c) 2D triangular element reconstruction mesh, and d) 3D tetrahedral element reconstruction mesh.

**VI. RESULTS**

In this section we show several 2D and 3D reconstructed images for a simulation, a phantom experiment and also some patient data.

**A. 2D Simulation: A blob-shape target**

Figure 5 shows the exact (a) and reconstructed (b) images of a simulation experiment with -100 dBm noise added at 1300 MHz. The first row of images corresponds to permittivity values, while the second row shows the map of conductivity values. The target had a random shape with dielectric properties of $\varepsilon_{\text{inc}} = 40.0$ and $\sigma_{\text{inc}} = 1.3$ S/m while the background medium was comprised of an 86:14 glycerin:water mixture with $\varepsilon_{\text{bk}} = 15.6$ and $\sigma_{\text{bk}} = 0.9$ S/m. In both reconstructed permittivity and conductivity image (Fig 5b), the target inclusion is successfully detected and the recovered values are very close to the exact property distributions (Fig 5a), though in the permittivity image, the shape of the target is more accurately obtained with fewer artifacts.

**B. 3D Phantom Experiment: Small Square-base Cylindrical Inclusion**

In order to study the effect of the target size on the reconstructed dielectric property distributions, a small square cylinder – of 1.0 cm side lengths and filled with a mixture of 60:40 glycerin and water – was immersed into an 80:20 glycerin and water bath. Figure 6 shows the extracted slices of the 3D reconstructed images at 1300 MHz – permittivity (top) and conductivity (bottom) – along with iso-surface values of $\varepsilon_r = 27.5$ and $\sigma = 1.35$ S/m. The true properties of the target inclusion and background medium were $\varepsilon_{\text{inc}} = 50.6$, $\sigma_{\text{inc}} = 1.28$ and $\varepsilon_{\text{bk}} = 22.4$, $\sigma_{\text{bk}} = 1.26$ S/m, respectively.

The target inclusion is clearly detected in both permittivity and conductivity images, although its shape is recovered more accurately in the permittivity image. In terms of finding the true dielectric property distributions of the target inclusion, the reconstructed conductivity values (~1.35 S/m) are more accurate than the reconstructed permittivity values (~28); nonetheless, it should be noted that the contrast between the target and the background values is much greater for permittivity than conductivity. In both images, there are some minor artifacts, especially at the boundary of the imaging domain, which may be due to the sources proximity.

![Phantom experiment with a square-base cylindrical inclusion](image)

**Fig. 6.** Phantom experiment with a square-base cylindrical inclusion: Extracted slices of the 3D reconstructed images at 1300 MHz – permittivity (top) with iso-surface values of $\varepsilon_r = 27.5$ and conductivity (bottom) with iso-surface values $\sigma = 1.35$ S/m.

**C. Patient Data**

Beside breast cancer detection, our clinical microwave imaging system at Dartmouth Hitchcock Medical Center (DHMC) is also used for therapy monitoring. We image patients with diagnosed breast tumors before and during
their therapy, and monitor how the tumor is affected by the treatment. Figure 7 shows the progress of a neo-adjuvant therapy monitoring patient from her first visit to 135 days afterwards, using the microwave imaging. The images were reconstructed in 2D and at 1300 MHz. (First row permittivity and second row conductivity images)

Based on radiologic studies and her clinical therapy reports, this patient had a complete response, in which all detectable signs of a tumor disappeared after the treatment. Our reconstructed images over the course of study agree with these clinical results and verify a significant tumor size reduction.

V. DISCUSSION

In this paper we showed how microwave imaging can be used for breast cancer detection and therapy monitoring. We performed simulation and phantom experiments to study the effect of non-uniform complex geometries and target size on our 2D and 3D image reconstruction algorithms. It was shown that despite the complexity and small size of the target inclusions, our image reconstruction algorithms are capable of not only detecting them, but also recovering their dielectric properties relatively accurately. We saw that the permittivity images are generally more precise in terms of finding the shape of the target, while the conductivity images can map true dielectric property distributions more correctly within the target region. Finally, we presented clinical patient data from a therapy monitoring case and imaged by our microwave imaging system at DHMC. In agreement with other standard radiological studies, our images showed that the size of the tumor decreased significantly during the course of treatment.

VI. CONCLUSION

Microwave imaging has the potential to become an appealing alternative and/or supplementary technique for breast cancer detection and therapy monitoring. Future work will involve improving our reconstruction algorithms to recover more accurate dielectric property distributions and detect even smaller tumors. We also plan to use our 3D reconstruction algorithm for patient data and we plan to obtain more detailed and accurate images.

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REFERENCES


