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Quantitative Microwave Imaging of Realistic Numerical Breast Phantoms Using an Enclosed Array of Multiband, Miniaturized Patch Antennas

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Abstract

We present a 3-D microwave breast imaging study in which we reconstruct the dielectric profiles of MRI-derived numerical breast phantoms from simulated array measurements using an enclosed array of multiband, miniaturized patch antennas. The array is designed to overcome challenges relating to the ill-posed nature of the inverse scattering system. We use a multifrequency formulation of the distorted Born iterative method to image four normal-tissue breast phantoms, each corresponding to a different density class. The reconstructed fibroglandular distributions are very faithful to the true distributions in location and basic shape. These results establish the feasibility of using an enclosed array of miniaturized, multiband patch antennas for quantitative microwave breast imaging.

Index Terms

Biomedical imaging; mammography; microwave imaging; multifrequency antennas; phantoms

I. Introduction

Microwave breast imaging via inverse scattering involves reconstructing the distribution of dielectric properties in the breast volume from measured scattered electromagnetic fields (see, for example, [1]–[3]). The significant dielectric properties contrast between different types of breast tissues [4] leads to differential scattering and thus allows for discrimination between tissue structures in a microwave image. Three-dimensional (3-D) microwave inverse scattering has potential benefits relative to current clinical imaging techniques for several applications in breast health and disease management, including breast density assessment (important because high breast density is a significant risk factor for breast cancer [5]), treatment monitoring, and screening in high-risk patients. Theoretical validations of 3-D microwave breast tomography using MRI-derived anatomically realistic

numerical breast phantoms, wherein the dielectric properties of the object to be imaged are known exactly, have been previously reported (e.g., [2] and [3]). However, this prior work has been limited to investigations of non-realizable arrays of idealized dipole antennas or line sources.

Microwave inverse scattering involves solving an inverse system that is ill-posed. The fidelity of microwave reconstructions is thus strongly influenced by the electric field spatial sampling density [6], the number of discrete frequencies used to find a solution [6], [7], and the degree to which noise or interference from outside the imaging region corrupts the measured electric fields. There is interest in designing antenna arrays that take these factors into account.

Fig. 1 shows our proposed array of multiband, miniaturized patch antennas surrounded by a conducting enclosure. The breast of the prone patient extends through the opening in the top panel of the array. An enclosed configuration is attractive, as it shields the imaging environment from outside interference and clutter. This shielding also simplifies the model of the array environment in the forward solution of the imaging algorithm. A few single-frequency enclosed arrays have been reported (e.g., [8] and [9]). However, the slot-loaded patch antennas comprising our array are designed to radiate at multiple resonant frequencies within the range relevant to microwave breast imaging while maintaining a relatively small physical footprint and thus a dense source population [10]. Arrays using this type of antenna are relatively straightforward to model on a Cartesian grid, which further simplifies the forward solution.

The proposed array fits within the interstitial space of an MRI patient-support platform and thus also allows for coregistration of microwave breast images with MRI [11]. This setup, together with a tissue-immobilizing thermoplastic mesh [12], allows microwave data to be acquired immediately following an MR scan while the breast remains stationary. This configuration permits a rigorous validation of microwave imaging as a viable clinical tool for breast imaging.

In this letter, we report a computational study of microwave imaging with our proposed array. Simulated scattered field data are acquired for heterogeneous, anthropomorphic testbeds of known dielectric distributions—namely, four MRI-derived numerical breast phantoms ranging in density from mostly fatty to extremely dense. We show that the reconstructed dielectric profiles are very faithful to the true dielectric profiles of the phantoms. This successful imaging of anatomically realistic breast phantoms demonstrates the feasibility of microwave breast imaging using an enclosed array of miniaturized, multi-band patch antenna.

II. Antenna Array

The microwave antenna array used in this study is in the form of a rectangular cavity. Conducting ground planes enclose the cavity on six sides. The ground plane of the top panel has an elliptical hole through which the pendant breast phantom descends. This hole conforms to the base of each breast phantom. The interior sides of the four lateral panels are

coated with a 60-mil substrate with dielectric properties $\epsilon_r = 6.15$ and $\sigma = 2.1$ mS/m. These properties are similar to those of Rogers RO4360 substrate.

The interior volume of the enclosed array is filled with an immersion medium whose dielectric properties are given by the Debye model parameters $\epsilon_\infty = 2.24$, $\epsilon = 0.73$, $\sigma_s = 0$, and $\tau = 5$ ps. This Debye model closely matches the dielectric dispersion of the biocompatible immersion medium of safflower oil over the frequency range 0.5–3.5 GHz [10].

Each of the four lateral array panels is populated with an eight-element subarray of slot-loaded patch antennas. The slot-loading approach is similar to that presented in [13]. The antennas are arranged in three rows, in a staggered 3-2-3 configuration that reduces the mutual coupling between vertically adjacent antennas [11], [14]. Vertical spacing and horizontal spacing between antennas (center-to-center) are both 48 mm. The patch antennas are of height 30 mm and width 28 mm. Slots are placed 2 mm from both radiating (i.e., top and bottom) edges of the patch antenna. The dimensions of the slots are 24×2 mm². The function of the slots is to move the lowest three transverse magnetic (TM) resonances of the antenna into the frequency range 0.5–3.5 GHz while maintaining a relatively small antenna footprint. The antenna feed is offset from the center of the patch by 3 mm in the vertical direction. The first three resonances of the antenna are at 1.60, 2.20, and 3.02 GHz. These resonances correspond to the TM₁₀₀, TM₂₀₀, and TM₃₀₀ mode, respectively. Note that the patch antenna has vertical polarization in these three modes. We determined the resonant frequencies of the antenna via finite-difference time-domain (FDTD) simulation.

III. Data Acquisition Simulation

The heterogeneous, anthropomorphic numerical breast phantoms used in this study are derived from MR scans of human subjects as described in detail in [15] (see [3] for modifications). We use one phantom from each of the four breast density classes as defined by the American College of Radiology, ranging from Class I, or mostly fatty, to Class IV, or extremely dense.

Data acquisition is performed by simulating each numerical testbed via FDTD formulated for simulating dispersive media described by single-pole Debye models, as in [3]. A graded mesh is used to model the substrate thickness with a fine discretization, while a coarser 1-mm discretization is used in the interior of the array, including the phantom region. The side and bottom boundaries of the computational domain are terminated with perfectly electrically conducting (PEC) planes. The top of the domain is terminated with a convolutional perfectly matched layer (CPML) absorbing boundary. An eight-cell buffer region is introduced between the CPML boundary and the PEC cap at the top of the array. The breast phantom is extended upwards through the elliptical hole into the eight-cell buffer region. The rest of the buffer is filled with free space.

The 32 antenna feeds are modeled as lumped-element resistive voltage sources. Each source sequentially transmits a modulated Gaussian pulse with a bandwidth covering 0.5–3.5 GHz. The resulting time-domain electric fields are recorded at all antenna feeds and converted to

phasor form at the TM_{100} , TM_{200} , and TM_{300} resonant frequencies via discrete Fourier transform.

IV. Imaging Technique

Estimates of the dielectric profiles of the numerical breast phantoms are computed using the multifrequency, Debye parameter formulation [3] of the distorted Born iterative method (DBIM) [16]. The forward solution is computed via FDTD on the same grid as was used for data acquisition. We avoid the inverse crime by adding Gaussian white noise to the simulated array measurements at a level corresponding to a signal-to-noise ratio (SNR) of 30 dB. SNR is defined here as the ratio between the total received signal power across all channels to the total noise power across all channels. As in [3], the inverse solution is constrained via the projected restart method such that the reconstructed Debye parameters are roughly proportional. We use the scalar formulation of the algorithm, in which the scattered electric field, incident electric field, and dyadic Green's function are replaced with their corresponding copolarized (in this case, vertical) components. The scalar approximation is only used for the purposes of the imaging algorithm; all FDTD simulations compute the full vector field quantities.

The DBIM requires computation of the heterogeneous background Green's function at each iteration of the algorithm. We calculate the Green's function from the incident field (which is computed in the forward solution of the algorithm) via the relationship

$$G^b(\mathbf{r}) = \frac{-jR_s E^i(\mathbf{r})}{\omega\mu_0 L V_s} \quad (1)$$

where E^i is the FDTD-computed incident electric field, G^b is the Green's function, L is the length of the lumped element resistive voltage source, V_s is the source voltage phasor, and R_s is the source resistance. Note that the computed Green's function takes into account multiple scattering within the enclosed array, as well as the characteristics of the antennas.

It is assumed that the dielectric properties of the breast voxels in the buffer region are unknown at each iteration of the DBIM algorithm. When the background profile is simulated in the forward solution, the row of voxels at the base of the breast phantom background profile is extended into the buffer region. The rest of the buffer region is again filled with free space.

V. Results

Figs. 2–5 show sequences of coronal cross sections through the true and estimated 3-D dielectric profiles for each phantom. In each case, the profiles are depicted in terms of Debye parameter ϵ . The three Debye parameters are highly correlated in both the true and estimated profiles, so ϵ_∞ and σ_s are not shown. Adjacent cross sections are separated vertically by 2 cm for the Class-I phantom (the largest of the four phantoms) and 1 cm for Classes II–IV.

The reconstructed profile is visually faithful to the true profile for each phantom. The low-level dielectric properties in the reconstructions correspond to fatty tissue. The reconstructed fibroglandular regions of tissues are accurate in location and basic shape. As in studies using idealized dipoles or line sources, the resolution is limited by the highest frequency used for the reconstruction. This contributes to the moderate blurring and underestimation of fibroglandular tissue dielectric properties in the reconstructions.

We also report results in terms of the performance metric $\cos(\phi)$ defined in [3], which quantifies the similarity in spatial distribution of dielectric properties between the actual and reconstructed profiles. The metric is given by

$$\cos(\phi) = \frac{\mathbf{p}^T \hat{\mathbf{p}}}{\|\mathbf{p}\|_2 \|\hat{\mathbf{p}}\|_2} \quad (2)$$

where $\mathbf{p}$ is the vector containing ϵ_∞ , ϵ , and σ_s at each voxel in the reconstruction region, and $\hat{\mathbf{p}}$ is the corresponding vector of reconstructed properties. Note that a perfect reconstruction would have $\cos(\phi) = 1$. For this study, we obtained $\cos(\phi) = 0.8157, 0.8408, 0.7972$, and 0.7447 , for the Class-I, Class-II, Class-III, and Class-IV phantoms, respectively. These values are comparable to the performance metrics for the reconstructions created by Shea *et al.* using an idealized dipole array (see [3, Fig. 8]). The somewhat diminished accuracy for the denser classes (III, IV) is likely due to higher attenuation associated with an increased percentage of fibroglandular tissue.

VI. Conclusion

This study presents results of imaging anatomically realistic, MRI-derived numerical breast phantoms using an enclosed array of miniaturized patch antennas and a 3-D multifrequency formulation of the distorted Born iterative method. Visual qualitative agreement between the true phantom profiles and the 3-D reconstructed profiles is evident in all coronal cross sections for all four phantom density classes. The performance metric $\cos(\phi)$ for each reconstruction is comparable to the corresponding metric for the idealized dipole array reconstructions despite using fewer and higher frequencies. Thus, an enclosed array of multiband patch antennas has excellent potential for microwave breast imaging.

The optimal choices for substrate, slot-loaded patch antenna, resonant frequencies, and configuration of antennas in each sub-array remain open questions. Potential improvements could be realized by using a slot-loaded patch antenna with more resonant frequencies in the 0.5–3.5-GHz range, though this may result in a loss of gain. Alternate array configurations may affect the radiation pattern inside the cavity, possibly leading to further performance improvement.

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Fig. 1. Illustration of a 32-element antenna array enclosing a Class-II numerical breast phantom. The substrates and ground planes on the four lateral panels are not shown to enable visualization of the interior. Axes are in centimeters.

Fig. 2. Class-I (mostly fatty) phantom: Coronal cross sections through the (*top*) true and (*bottom*) reconstructed profiles for the Debye parameter ε . Adjacent cross sections are separated vertically by 2 cm.

Fig. 3.

Class-II (scattered fibroglandular) phantom: Coronal cross sections through the (*top*) true and (*bottom*) reconstructed profiles for the Debye parameter ε . Adjacent cross sections are separated vertically by 1 cm.

Fig. 4.

Class-III (heterogeneously dense) phantom: Coronal cross-sections through the (*top*) true and (*bottom*) reconstructed profiles for the Debye parameter ε . Adjacent cross sections are separated vertically by 1 cm.

Fig. 5. Class-IV (extremely dense) phantom: Coronal cross sections through the (*top*) true and (*bottom*) reconstructed profiles for the Debye parameter ϵ . Adjacent cross sections are separated vertically by 1 cm.