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Hybrid Microwave Tomography Technique for Breast Cancer Imaging

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Abstract - **This paper addresses a two-dimensional inverse scattering method with a combination of tomography and radar methods for breast cancer detection. In order to rapidly construct high resolution images displaying the location, size, permittivity and conductivity of malignant tumors inside the body, the collected reflection from the scattered fields present in the scan area is segmented and their associated dielectric property maps are calculated. The dielectric profiles are obtained by using a technique that combines frequency domain finite difference time domain $(FD)^2TD$ analysis with genetic algorithm (GA) optimization. The applications of the proposed method can vary from medical imaging to nondestructive testing of materials and structures. The proposed technique yielded promising results when applied to simulated data.**

I. INTRODUCTION

IN many parts of the world, breast cancer is the leading cause of morbidity and mortality among women and it is a major cause of cancer death, next only to lung cancer. As with many other types of cancer, early detection and medical intervention is a fundamental factor in the long-time survival and quality of life of patients. Currently, X-Ray mammography is the most commonly used technique for detection of early stage non-palpable breast tumors. Despite several efforts to improve the quality of the images produced by this modality, several limitations persist. Due to the low contrast between cancer and normal tissue observed in X-ray imagery, 70% of breast lesions detected by mammography turned out to be benign while 4% to 34% percent of breast tumors are missed by this imaging modality [1]. Another factor to consider is the pressure on the affected area during data acquisition, making it a stressful and uncomfortable process. In recent years, techniques such as Magnetic Resonance Imaging (MRI) and Ultrasound have been successfully used as an alternative to mammography for Breast Cancer Detection (BCD) purposes. However, MRI is expensive and ultrasound does not have spatial resolution that is sufficiently accurate to be considered for mass screening. During the last 10 years, Microwave based methods have been proposed as an alternative for early breast cancer detection. These techniques exploit the scattering produced by the differences in dielectric properties between breast and cancer tissue at the microwave frequency range. During the last few years, some active microwave approaches, such as Microwave Tomography (MT) [2] and Ultra Wide Band Microwave Radar (UWBMR) [3], have been proposed.

As discussed in [3] and [4], UWBMR techniques use the reflections from the scatterers present in the scanned area in order to determine their spatial location and dimensions. On the other hand, MT techniques try to recreate the dielectric properties map of the scan area from measurements of the microwave energy transmitted through it. Although promising results have been obtained by these two approaches in both real and simulated data, MT and UWBMR techniques still present limitations that could restrict their use in a clinical environment. UWBMR methods do not provide a dielectric map of the collected scattered field responses and MT approaches are computationally expensive. The purpose of this paper is to propose a novel method of producing accurate dielectric maps in order to distinguish between malignant and other breast tissue and with a low computational cost. The proposed technique extracts the spatial location and dimension from the most prominent scatterers in UWBMR imagery in order to determine their dielectric properties using nonlinear inverse scattering techniques [5].

This paper is organized as follows: In the first part of this paper methodology is explained. Next, the results of applying the dispersive characteristic of breast tissue to the forward simulation of a breast phantom are discussed. This procedure is illustrated using a two-dimensional breast model, consisting of various proportions of skin, fatty breast tissue and malignant lesion. We have developed a 2-D $(FD)^2TD$ model that incorporates dispersion using the first-order Debye equation in the Finite Difference Time Domain (FDTD) model. Utilizing this method, the behavior of certain parameters for various types of biological tissues in the frequency range of 1-12 GHz is studied for the purpose of breast cancer imaging. In the next part, ultra-wide band microwave radar as a spatial method for reconstruction, and GA optimization for obtaining a map of dielectric properties are discussed. Finally, the simulated results of this novel method are presented.

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II. METHODOLOGY

Fig. 1 depicts a cross-sectional view in the x-y plane of a breast phantom under test. The goal is to image a non-homogenous object (tumor) inside a homogenous background (normal tissue). The homogenous background is covered by an outer layer (skin) with different dielectric properties. This object is illuminated by a transverse magnetic to z-axis (TM_z) plane wave. The plane wave impinges on the object and the scattered field is collected by the receiving antennas that surround the object. The collected scattered field is used as a source of data for the radar reconstruction method. The collected reflections are focused using a wavefront reconstruction approach [6]. First, noise is removed and then the resulting image is segmented using the method proposed by the authors in [7]. The target signatures are labeled and their spatial locations and dimensions are stored for the following steps. Next, the FDTD/GA method [5] is used to obtain the exact location and dielectric properties of the object (malignant tumor). FDTD/GA works as follows: The numerical algorithm starts with a set of random initial guesses of the structure. Then a full wave analysis using FDTD is carried out for each member of this set to find the values of scattered fields for each case. The calculated scattered field is compared with the measured field and a GA optimization is used to find the structure that minimizes the difference between the simulated and measured scattered fields.

III. $(FD)^2TD$ SIMULATION

FDTD is a computationally intensive method that we used to solve Maxwell's equations in a two-dimensional space for TM_z waves. One benefit of the time domain approach is that it gives a broadband output from a single execution of the program [8]. However, the main reason for using the FDTD approach is its effectiveness as a technique for calculating electromagnetic fields in multilayer inhomogeneous objects such as biological tissues. With a large number of unknown parameters related to the object under test, the FDTD approach outpaces other methods in efficiency and provides accurate results of field penetration into biological tissues. The frequency dependence of biological materials can be efficiently described in the time domain using the standard Debye or Lorentz models [9]. We derived a Debye first-order model given by (1), where ε_0 is the permittivity of the free space, ε_s and ε_∞ are the dielectric constants at zero (static) and infinite frequency, respectively, ω is angular frequency and τ_0 is the relaxation time constant. $(FD)^2TD$ is an extended version of the conventional FDTD that can handle dispersive materials. In order to simulate the breast of a patient in prone position, an infinitely extended cylinder in z-direction is considered as the structure under study (Fig.1).

The receiver antennas are spread around the breast phantom to measure scattered fields in the far field zone. A plane wave impinges on the structure at different angles and passes through it. Then, the scattered fields are accumulated by the receiver antennas around the structure.

In FDTD methods, Maxwell's equations are solved in a closed area. This means that the solution area for scattering problems in an infinite space, such as the one that we are dealing with, needs to be truncated by an absorbing boundary condition (ABC). We chose to use the Perfectly Matched Layer (PML) which is a very efficient ABC. PML ABC is based on an artificial absorbing layer surrounding the simulation region. In order to assign differing losses to individual directional components, this method decomposes the fields in the boundary region, [8].

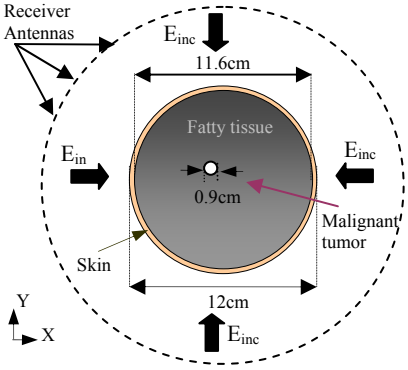

Fig. 1. Breast phantom with skin, breast tissue and a malignant tumor at an off-center location (-0.75, 0.75 cm) with Debye parameters shown in Table I (top view)

A. *Dispersive biological tissue simulation*

Microwave imaging is based on the contrast between the dielectric properties of materials. As this statement indicates, the most important way to have accurate images in this method is to consider the frequency dependence of dielectric characteristics of biological tissue at the microwave range. Fig. 2 shows the dielectric properties of fatty and malignant tissues with Debye parameters specified in Table I at different frequencies.

$$\varepsilon_r(\omega) - j\frac{\sigma(\omega)}{\omega\varepsilon_0} = \varepsilon_\infty + \frac{\varepsilon_s - \varepsilon_\infty}{1 + j\omega\tau_0} - j\frac{\sigma_s}{\omega\varepsilon_0} \quad (1)$$

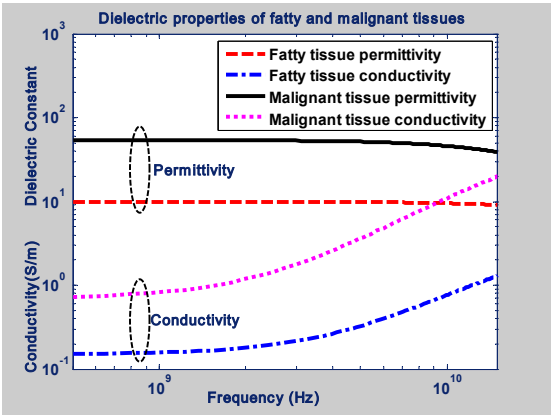

Fig. 2. Frequency variation of electrical properties for fatty and malignant tissues.

One can see that there is approximately 6% permittivity difference between fatty tissue at high (12GHz) and low (1GHz) frequencies and this value goes up to about 23% for

malignant tissues. The conductivities of these two tissue types are very different. In order to consider the dispersive characteristics of the breast tissues, the $(FD)^2TD$ model was developed. Our calculations show that the penetration depth is decreased by increasing the frequency. For frequencies beyond 15GHz, the penetration depth becomes a practical problem.

IV. WAVEFRONT RECONSTRUCTION

Consider the scan pattern formed by a circular array of radius *Z,* formed by *N* antennas, as shown in Fig. 1. For the following discussion, *T* point scatterers are assumed to be located inside the area enclosed by the array. Also, the center of the antenna array is considered to be the origin of the coordinate system and a polar coordinate system is used in order to simplify the calculations. In this scenario, the location of the *p* scatterer located at (x_p,y_p) is denoted by (r_p, ϕ_p) where $r=\sqrt{x_p{}^2+y_p{}^2}$ and $\phi=\tan^{-1}(y_p/x_p)$. Also, the distance between the n^{th} antenna and the *p* scatterer is given by $R'=\sqrt{Z^2+r_p{}^2-2\cdot Z\cdot r_p\cos(\phi_p-\theta_n)}$, where (Z,θ_n) are the polar coordinates of the n^{th} antenna.

As the scan process is performed, a signal $f(t)$ is irradiated from each of the n^{th} array elements and the received reflections for the n^{th} element can be expressed as:

$$s(t,\theta_n)=\sum_{j=1}^{T}\sigma_j f\left(t-\frac{2\sqrt{Z^2+r_j{}^2-2\cdot Z\cdot r_j\cos(\phi_j-\theta_n)}}{\nu}\right) \quad (2)$$

where ν is medium propagation speed.

As discussed in [3], the obtained range profiles form nonlinear signatures due to the difference in travel times. This fact makes it difficult to accurately determine the dimensions of each target present in the scanned area.

In order to properly visualize the dimensions and locations of the actual targets, the collected data must be focused [4]. When the Fourier transform is used in the signal travel direction, a set of Phase Modulated (PM) signals is obtained [10]. When observed from the different scan positions, the instantaneous frequency of this signal varies as the antenna is closer to the target. In order to take advantage of this phenomenon, the Fourier transform is calculated along the scanning trajectory direction using the stationary phase method [11]. For each of the previously discussed acquisition trajectories, the frequency responses are given by:

$$S_c(\omega,\varepsilon)=\sigma_c F(\omega,\varepsilon)\cdot e^{-i\left(\sqrt{4k^2r_p{}^2-\gamma^2}+\sqrt{4k^2Z^2-\gamma^2}+\sin^{-1}\left(\frac{\gamma}{2kZ}\right)+\sin^{-1}\left(\frac{\gamma}{2kr_p}\right)-\pi\right)} \quad (3)$$

where γ and ω are the frequency counterparts of θ and t respectively, and *k* denotes the wavenumber in free space**.**

Given that the frequency response of a point scatterer at (x_n, y_n) has the form $\exp(-ik_y y_n - ik_x x_n)$, the goal is to obtain similar a response for each target by correcting the phase variations of the collected signals. This is done by multiplying the frequency response by the complex conjugate of the PM terms causing the artifacts. As explained in [6] the phase distortion is caused by $\exp(-i\cdot(\sqrt{4k^2Z^2-\varepsilon^2}+\sin^{-1}(\gamma/2kZ)+\sin^{-1}(\gamma/2kr_p)-\pi))$. Next, the compensated data needs to be transferred from the frequency-wavenumber domain to the spatially frequency domain. This is done by using the mapping function $(k_x,k_y)=(k\cos\theta_n, k\sin\theta_n)$. However, due to the non-linear nature of the mappings, the resulting spatial frequency spectrum is unevenly spaced [10]. In order to use the standard Fourier transform, an interpolation process is performed on the focused responses to obtain an evenly sampled spectrum. Finally, the inverse double Fourier transform of the data produced in the previous step is calculated to properly display the focused data in the spatial domain.

V. GA OPTIMIZATION

The GA is a popular evolutionary global optimization method that performs very well for problems that involve a high number of parameters and high degrees of non-linearity. In a GA, a set or population of potential solutions (individuals) evolves toward a global minimum. A fitness function is evaluated for every individual to check how good it is. The good solutions are used to create the next set of possible solutions (used as parents for the next generation). In this research, the dielectric properties and location of the solution domain are used as parameters to optimize a cost function. After discretizing the solution space - that the radar method has identified a priori - into a number of small patches, each patch is assigned a dielectric permittivity and conductivity pair (ε_j,σ_j), where *j* is the index for the patch location. To speed up the optimization process and include a priori information about the materials, a look up table (Table I) of dielectric properties of possible materials is created. Each individual is a set of possible dielectric permittivities and conductivities assigned to each point in space. The fitness function is the difference between the simulated and collected scattered fields at the observation points (for 10 frequencies between 1-10 GHz). The FDTD method is employed to compute the scattered electric field at the observation points to provide information needed at each generation of the GA optimization procedure. After reaching the maximum number of generations, the best individual is chosen as a final image which provides dielectric values of each location. The minimum resolution for this method is dependent on the size of patches that we divided the search space into, and therefore, there is a trade-off between the size, resolution and run time.

TABLE I: DEBYE PARAMETERS FOR THE DISPERSIVE MATERIALS INCLUDED IN THE ELECTROMAGNETIC MODEL [12].

Medium	ε_∞	ε_s	σ_s	τ_0
Skin	4.00	37.00	1.10	7.23e-12
Tumor	3.99	54.00	0.70	7.00e-12
Muscle	4.00	54.00	0.20	7.23e-12
Fatty breast tissue	7.00	10.00	0.15	7.00e-12

VI. SIMULATION RESULTS

In this section, the effectiveness of the hybrid method of tomography and radar reconstructed method for early breast cancer detection is discussed. Fig. 1 illustrates the multilayer breast phantom used in this work that is considered to be formed of skin, normal breast tissue, and malignant tumor tissue. The model includes a 2 mm thick skin layer with an 11.6 cm diameter circle for a normal breast and a 0.9cm diameter circular malignant tumor at an off-center location (-0.75, 0.75 cm) . The FDTD model uses a uniform grid with square unit cells. The cell side in both x and y direction are 0.3 mm in the highest-resolution simulation and a time step of 0.5 ps. The incident wave was a Gaussian pulse $E(t) = \exp(-(t-t_0)^2/2T^2)$ where $t_0=60\Delta t$ and $T=10\Delta t$ and 10 layers of PML absorbing boundary condition was used. The Debye parameters of dielectric properties of skin, breast and tumor are calculated according to Table I. There are 72 receiver antennas located at 64cm away from the skin. A GA optimization program ran for 30 generations, each with a population of 10. On a 3GHz personal computer with 1GB RAM, the runtime for a conventional GA is about 70 hours. With 4 computers running the GA program in parallel the runtime reduces to 20 hours. Fig. 3 shows the result of the radar reconstruction method and Fig. 4 shows the reconstructed image with exact locations and dielectric properties.

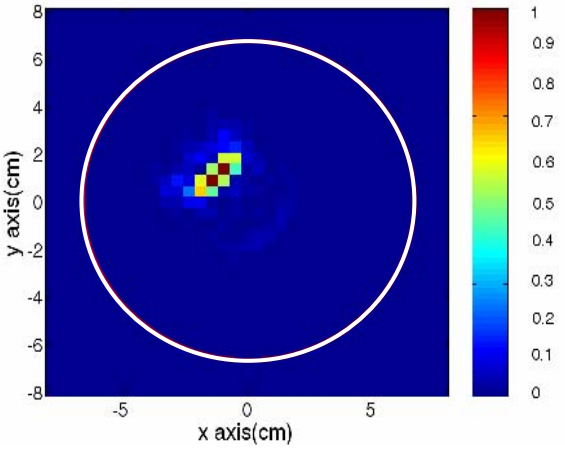

Fig. 3. Reconstructed image obtained using radar technique

VI. CONCLUSION

A novel technique for the calculation of dielectric maps in breast microwave scans was presented in this paper. The proposed method determines the position and dimension of the most prominent reflections present in focused UBWMR imagery in order to reduce the computational cost associated with the non-linear backscattering methods that are used to determine the dielectric properties of such scatterers. The proposed method was tested on a computational model that mimics the dielectric properties of breast, skin, and cancer tissues, yielding promising results.

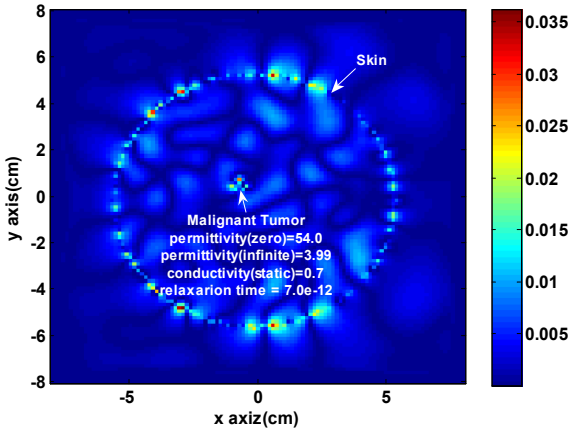

Fig. 4. Reconstructed image of a 0.9cm-diameter malignant tumor inside a breast phantom

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