An Ultrawideband Microwave Imaging System for Early Detection of Breast Cancer

By
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AN ULTRAWIDEBAND MICROWAVE IMAGING SYSTEM FOR EARLY DETECTION OF BREAST CANCER

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Motivated by the critical need for complementary and/or alternative modalities to X-ray mammography for early stage breast cancer detection, we have proposed a method of ultrawideband microwave imaging for detecting backscattered energy from small malignant breast tumors. This thesis presents a detailed numerical and experimental investigation of the proposed microwave imaging system.

In the proposed system configuration, each antenna in the array sequentially transmits a low-power UWB signal into the breast and records the backscatter. The backscatter signals are passed through a beamformer, which spatially focuses the waveforms to image backscattered energy as a function of location in the breast. First a simple beamforming method using basic time-shift-and-sum beamforming algorithm is proposed. Our 2-D and 3-D numerical studies have demonstrated the feasibility of detecting backscattered energy from small malignant breast tumors using this straightforward scheme without solving the inverse problem. Improved algorithms for both removing artifact components and space-time beamforming are introduced later. The robustness of these algorithms is investigated using extensive numerical studies.

The preliminary experimental investigation of the microwave imaging system is based on multilayer simple breast phantom consisting of a homogeneous normal
breast tissue simulant covered by a thin layer of skin simulant. A small synthetic malignant tumor is embedded in the breast phantom. We have developed several tumor simulants that yield the range of dielectric contrasts between normal and malignant tissue that are expected in clinical scenarios. A microwave sensor comprised of a planar synthetic array of compact ultrawideband (1-11 GHz) antennas is used to transmit and receive microwave energy. Small (< 5-mm) synthetic tumors with malignant-to-normal dielectric contrasts down to 1.5:1 are successfully imaged. Our experimental results suggest that microwave imaging via space-time beamforming offers the potential of detecting small breast tumors using state-of-the-art but readily available hardware and robust signal processing algorithms.

I certify that I have read this thesis and certify that in my opinion it is fully adequate, in scope and in quality, as a dissertation for the degree of Doctor of Philosophy.

Susan C. Hagness Date
Assistant Professor of Electrical and Computer Engineering
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In the memory of my father.
I would like to extend my deepest appreciation and gratitude to the faculty at University of Wisconsin – Madison. In particular, I want to thank my advisor Susan C. Hagness, for her guidance not only on my research project, but throughout my graduate program. She has been an extraordinary advisor, mentor, and role model to me. I also want to thank Prof. Barry D. Van Veen and Prof. Daniel W. van der Weide from the Department of Electrical and Computer Engineering. They have brought many insights and new ideas to my research project. My discussions with them have been extremely helpful.

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Chapter 1

Introduction

1.1 Background and motivation

Breast cancer is one of the leading causes of death among women in United States. More than 180,000 new cases of invasive breast cancer are diagnosed and more than 40,000 deaths result from the disease each year [1]. Early detection and timely medical intervention are key factors affecting long-term survival and life quality of breast-cancer patients.

Mammography, which is the X-ray imaging of a compressed breast, remains the primary screening method for detecting non-palpable early-stage breast cancer. However, despite significant progress in improving mammographic technique, well recognized limitations persist [1]. Approximately 4%-34% of all breast cancers are missed by conventional mammography [2] while nearly 70% of all breast lesions identified by mammography turn out to be benign [3]. The unsatisfactory sensitivity and specificity to a large extent arise from the small (a few percent) intrinsic contrast between the radiographic density of malignant and normal tissue exploited by X-ray, particularly in premenopausal women with radiographically dense breast tissue. Furthermore, since X-ray mammography is a 2-D projection imaging technique, breast compression is required to create a uniform volume of tissue between the source and receiver located on opposing sides of the breast. This
uncomfortable breast compression, along with concerns of accumulating low-dose ionizing radiation over repeated scans, may reduce patient compliance with the screening recommendations.

Other breast imaging methods approved by the Food and Drug Administration (FDA) include magnetic resonance imaging (MRI), ultrasound, scintimammography, thermography, and electrical impedance imaging [1]. These alternative modalities have shown potential to improve diagnosis in certain cases when used as a complement to X-ray mammography. However, because of limitations in image quality, diagnosis accuracy, or availability, these methods are not yet used routinely in screening tests. Additional clinical trials are needed to assess the sensitivity and specificity of these procedures.

The limitations of existing breast imaging modalities motivate the search for alternative breast screening tools that image other physical tissue properties or metabolic changes. According to [1], an ideal breast screening method is characterized by the following:

- has low health risk
- is sensitive to tumors and specific to malignancies
- detects breast cancer at a curable stage
- is noninvasive and simple to perform
- is cost effective and widely available
- involves minimal discomfort, so the procedure is acceptable to women
- provides easy to interpret, objective, and consistent results
One of the alternative modalities under investigation for breast cancer detection is microwave imaging. This modality avoids using ionizing radiation and breast compression, resulting in safer and more comfortable exams. Microwave imaging systems are also expected to be of relatively low-cost and therefore can be available to the majority of the public. Both passive and active microwave imaging techniques are being researched for breast cancer detection [4]. Passive microwave radiometry [5], [6] exploits temperature differences between malignant and normal breast tissue due to elevated metabolism in fast-growing malignant tumors. Active microwave imaging exploits the dielectric contrast between malignant tumors and normal breast tissue at microwave frequencies. We are particularly interested in an active microwave approach since we believe it has the potential to improve sensitivity and specificity of the screening test due to the significant malignant-to-normal breast tissue dielectric contrast, as suggested by published measurements and our own preliminary measurements conducted on freshly excised breast tissue.

1.2 Dielectric properties of breast tissues

As stated in Sec. 1.1, the most important underlying rationale of active microwave imaging for breast cancer detection is the dielectric-properties contrast between normal and malignant tissue at microwave frequencies. This section briefly reviews several published dielectric measurements on breast tissues in the radio/microwave frequency range.

Chaudhary et al reported the dielectric properties of 5-mm samples of excised normal and malignant breast tissue for 15 patients in different age groups [7].
Their lower frequency measurements (3-100 MHz) were conducted using a 250-A RX-meter, while their higher frequency measurements (100 MHz-3 GHz) were based on time domain spectroscopy. Surowiec et al [8] published dielectric properties measurements for breast carcinoma and surrounding nonmalignant breast tissue over the 20 kHz to 100 MHz frequency range. Those measurements were conducted using a vector network analyzer. Both groups observed from the measurements that the permittivity and conductivity of malignant breast tissue were considerably higher than those of normal breast tissue. More recently, Joines et al measured the dielectric properties of normal and malignant tissues over 50 to 900 MHz from different part of human body using a vector network analyzer [9]. Their measurements indicated a contrast ratio of about 3:1 for permittivity and 7:1 for conductivity between malignant and normal breast tissue. Most of the above measurements were conducted using open-ended coaxial probe techniques. Jossinet [10] measured the impedivity of 64 freshly excised breast tissue specimens at very low frequencies (488 Hz-1 MHz). Statistical analysis showed significant differences between cancerous and non-cancerous tissue [11].

Campbell and Land [12] measured the dielectric properties of breast tissue at 3.2 GHz using a resonant cavity perturbation technique. Their measurements do not agree with the work cited above in that they are inconclusive about the contrast between normal and malignant breast tissues. These discrepancies may arise because of their experimental protocol. For example, their measurement technique involved cutting and pushing the tissue specimen into a sample holder, which may cause fluid loss in high-water-content tissue and increase the density of low-water-content tissue. Also, air gaps may have been introduced in the sample test chamber.
Paulsen et al indirectly measured the dielectric properties of normal breast tissue using a clinical prototype of a microwave tomographic system operating at 900 MHz [13]. Their reconstructed average permittivity and conductivity profile of normal breast tissue were considerably higher than the previously published values measured by open-ended coaxial probes, suggesting that the contrast between normal and malignant breast tissue may be closer to 2:1.

In summary, most of the measurement data sets suggest that there indeed exists a significant contrast between the dielectric properties of malignant and normal breast tissue at frequencies below 3 GHz. In addition, the variability in the dielectric constant, $\epsilon_r$, and conductivity, $\sigma$ appears to be no greater than $\pm 10\%$ [7], [9]. The dielectric properties of malignant tumors show no significant variation with tumor age [14], suggesting that the large contrast exists at the earliest stages of tumor development. The enhanced dielectric properties of breast carcinomas appear to arise in part from increased protein hydration [15]. The contrast is further enhanced by the vascularization of malignant tumors. As a result, malignant tumors have large microwave scattering cross-sections relative to comparably sized heterogeneity in normal breast tissue.

In order to extrapolate breast-tissue dielectric parameters to above 3 GHz, a first-order (Debye) dispersion formulation is used to model the frequency dependence of $\epsilon_r$ and $\sigma$ (S/m):

$$\epsilon_r - j \frac{\sigma}{\omega \epsilon_0} = \epsilon_\infty + \frac{\epsilon_s - \epsilon_\infty}{1 + j \omega \tau} - j \frac{\sigma_s}{\omega \epsilon_0}. \quad (1.1)$$

Here $\epsilon_s$ is the static relative permittivity, $\epsilon_\infty$ is the relative permittivity at infinite frequency, $\sigma_s$ is the static conductivity, and $\tau$ is the relaxation time.

Figure 1 shows the results of curve-fitting the Debye equation to published
Figure 1: Single-pole Debye curve fits of measured baseline dielectric-properties data for normal and malignant breast tissue at radio and microwave frequencies.

data in [7], [8], and [9]. For normal breast tissue, the Debye parameters ($\epsilon_s = 10$, $\epsilon_\infty = 7$, $\sigma_s = 0.15$ S/m, $\tau = 7.0$ ps) yield $\epsilon_r = 9.8$ and $\sigma = 0.4$ S/m at 6 GHz, while the Debye parameters for malignant breast tissue ($\epsilon_s = 54$, $\epsilon_\infty = 4$, $\sigma_s = 0.7$ S/m, $\tau = 7.0$ ps) yield a dielectric constant of 50.7 and conductivity of 4.8 S/m at 6 GHz. This data extrapolation process suggests a contrast between malignant and normal breast tissue on the order of 5:1 in dielectric constant and 12:1 in conductivity over the microwave frequency range.

Thus, the dielectric properties contrast at microwave frequencies appears to be much larger than the few-percent contrast exploited by X-rays. Consequently, microwave imaging has the potential to provide improved sensitivity over X-ray
mammography for early-stage breast cancer detection. Microwave attenuation in normal breast tissue is low enough to make signal propagation quite feasible even through large breast volumes.

1.3 Active microwave imaging methods under development

Three types of active microwave breast imaging techniques have been proposed: hybrid microwave-induced acoustic imaging, microwave tomography, and ultrawideband (UWB) microwave radar techniques. In the hybrid approach [16], [17], microwave signals are transmitted into the breast to selectively heat tumors, and ultrasound transducers are used to detect pressure waves generated by tumor expansion. This method is based on the different heat absorption rate due to the conductivity contrast between malignant and normal breast tissue.

Non-hybrid methods involve illuminating the breast with microwaves and then measuring scattered microwave signals. The received waveforms are used to infer the tissue dielectric-properties distribution inside of the breast. The goal of classical microwave tomography is to recover the dielectric-properties profile of the breast by solving an inverse problem. In tomographic microwave imaging approaches [18], [19], [20], [13], [21], several microwave transmitters illuminate the breast, and scattered fields in numerous locations are measured. The spatial distribution of the tissue dielectric properties are obtained from the transmitted (incident) and scattered (received) fields. Promising initial clinical results for breast imaging have been obtained recently [13]. The challenge of microwave tomography, however, is that it involves the solution of an ill-conditioned nonlinear
inverse-scattering problem which is often computationally intensive and is inherently limited by vulnerability to small experimental uncertainties and noise. Other problems such as non-uniqueness of solutions, resolution, object size and geometry, computational resource requirements, and amount of a priori information need to be successfully addressed in microwave tomographic imaging.

Hagness et al proposed an alternative approach based on ultrawideband (UWB) radar techniques [22], [23]. Analogous to ground penetrating radar [24], UWB radar techniques illuminate the breast with an ultrawideband pulse from a number of physical antenna locations and collect the backscattered signals. In contrast to microwave tomography which attempts to completely reconstruct the dielectric-properties profile, this approach seeks only to identify and locate scattering sites which arise from the significant dielectric contrasts between normal breast tissue and malignant lesions.

Two UWB radar configurations are currently under investigation. Fear and Stuchly have evaluated a cylindrical system configuration [25], [26], [27], [28]. In this configuration (Fig. 2), the patient is oriented in a prone position with the breast naturally extending through a hole in the examination table. The ultrawideband antenna elements are distributed around the breast to create a cylindrical array. In a second configuration—the planar configuration, the patient is oriented in a supine position and the ultrawideband antenna array is placed close to the surface of the naturally flattened breast (Fig. 3). An important feature of this configuration is the capability to access the region close to the chest wall and upper outer quadrant of the breast where almost 50% of cancers occur [29]. This is the configuration addressed in this dissertation.
Figure 2: Patient orientation for the cylindrical array configuration.

Figure 3: Patient orientation for the planar array configuration.
1.4 Research objectives and outline

This dissertation presents a numerical and experimental investigation of the UWB microwave imaging system for detecting small breast tumors. The primary goals of this research include:

- the characterization of UWB microwave signal propagation in breast tissue and scattering from malignant tumors
- the development and evaluation of signal processing algorithms to detect small breast tumors
- the design and characterization of an UWB antenna element
- the demonstration of the feasibility of system implementation using an initial experimental setup and simplified breast phantoms

In our currently proposed system, each antenna in the array sequentially transmits a low-power UWB signal into the breast and records the backscatter. The backscatter signals are passed through a beamformer, which is designed to image backscattered energy as a function of location in the breast. The goal of the signal processing of the UWB approach is to detect scatterers by spatially focusing received backscatter waveforms. Chapter 2 proposes a simple beamforming method where the backscattered waveforms are synthetically focused by a basic time-shift-and-sum beamforming algorithm [30]. Our 2-D and 3-D numerical studies have demonstrated the feasibility of detecting backscattered energy from small malignant breast tumors using this straightforward scheme without solving the inverse problem [30] [27]. In Chapter 3, improved algorithms for both removing artifact components and space-time beamforming are introduced [31] [32]. The
robustness of these algorithms is investigated using extensive numerical studies. Chapter 4 presents the numerical and experimental study of an UWB antenna element suitable for the UWB microwave breast imaging application. Finally, a first-generation experimental demonstration of the space-time imaging method is provided in Chapter 5. Simplified experimental breast phantoms have been developed to show the experimental feasibility of the imaging system.

The finite-difference-time-domain (FDTD) [33] method is used intensively in this research to simulate microwave interactions with breast tissue, to generate representative backscatter waveforms, and to optimize antenna performance and array arrangement. Some of the FDTD models include complicated geometry, such as those involving anatomically realistic breast models. Descriptions of these simulations are provided in each chapter when the model is first introduced.
Bibliography


Chapter 2

Microwave Imaging via

Time-Shift-and-Sum

Beamforming

Spatial focusing of the backscattered waveforms can be achieved with a simple time-shift-and-sum beamforming approach. In this approach, a simple calibration step is applied to the backscatter waveforms received at all antenna locations to remove the early-time artifact components, then they are time-shifted and added to create a synthetic focal point. The position of the focus is scanned throughout the breast by adjusting the distribution of time shifts of the stored backscatter waveforms for each new focal point. If a high-contrast scattering object, such as a malignant tumor, exists at the focal point, the waveforms add coherently. Clutter signals generated by the heterogeneity of normal breast tissue surrounding the focal point will add incoherently. In this manner, tumor backscatter signals are enhanced while clutter signals are suppressed. Systematic scanning of the synthetic focus from point to point within the breast creates a microwave image of significant scattering points within the breast.

To test the image reconstruction algorithm, we developed 2-D anatomically realistic FDTD breast models [1] and 3-D simplified FDTD breast models [2].
The time-shift-and-sum beamforming algorithm is applied to FDTD-computed backscatter signals, resulting in microwave images that clearly identify the presence and location of the malignant lesion. These simulations demonstrate the feasibility of detecting and imaging small breast tumors using this straightforward approach.

2.1 Time-shift-and-sum beamforming: A simple synthetic focusing algorithm

The backscattered waveforms recorded at the antenna locations include early and late time content. The early-time response is dominated by the incident signal, skin backscatter and antenna reverberation. The late-time response contains the tumor response and clutter due to the natural heterogeneity of the breast. The signal processing goals are to reduce the early-time content, which is of a much greater amplitude than the tumor response, and to selectively enhance the tumor response while suppressing the clutter to permit reliable detection of tumors. Using a straightforward algorithm, images are reconstructed with the post-processed signals. The image formation steps are described below.

1) Calibration

The purpose of the calibration step is to remove the incident pulse, skin backscatter and antenna reverberation from the recorded waveforms. A reference waveform is created by averaging the $M$ waveforms recorded at various antenna locations based on the assumption that these waveforms have similar artifact content. The reference waveform is then subtracted from each of the $M$ original backscatter waveforms, resulting in $M$ calibrated backscatter waveforms that essentially contain only the tumor response and clutter signals. As a final step in the
calibration process, low-frequency signal content is removed by subtracting from each waveform its moving average.

2) Image reconstruction

The calibrated data set is in the form of $M$ vectors: $A_1, A_2, ..., A_M$, with each vector containing $N$ time-sampling points. The following algorithm describes the coherent-summing process. The $m^{th}$ discrete-time delay needed to achieve a synthetic focus of the $M$ backscatter waveforms at position vector $\vec{r}$ in the breast is given by $\tau_m(\vec{r}) = 2d_m(\vec{r})/(v\Delta t)$, where $d_m(\vec{r}) = |\vec{r} - \vec{r}_m|$ is the distance between the synthetic focal point and the $m^{th}$ transmit/receive antenna element located at position $\vec{r}_m$, $v$ is the average velocity of propagation in the breast at the center frequency of the pulse, and $\Delta t$ is the time-sampling interval between data points. In 2-D space, $\vec{r} = (x, y)$; in 3-D space, $\vec{r} = (x, y, z)$.

The UWB transmitted pulse is assumed here to be a differentiated Gaussian pulse with a temporal duration of approximately 100 ps. Because this excitation signal has a zero-crossing at its center point in time, the backscattered signal also has a zero-crossing at a time delay corresponding to the round-trip distance between the antenna and the focal point coinciding with the scattering location. To obtain a non-zero sum in the subsequent step, each calibrated backscatter waveform is integrated over time to obtain $M$ vectors, $B_1, B_2, ..., B_M$, so that the peaks of the waveforms coincide with the focal point location, allowing for the coherent addition of local maxima via straightforward time-shifting.

Compensation for radial spreading and/or path loss is applied to the signals. Radial spreading correction accounts for the decrease in amplitude of an expanding spherical wave, while path loss compensation corrects for the reduction in signal strength due to propagation through lossy breast tissue. Radial spreading can be
approximated with a $1/r$ model in a 3-D system and $1/\sqrt{r}$ model in a 2-D system, where $r$ is the distance from the antenna to the focal point. The approximate path loss factor can be derived from the average conductivity of the normal breast tissue. Alternatively, these compensation factors can be obtained through FDTD simulation. In the FDTD simulation, the ultrawideband antenna is immersed in a homogeneous lossy medium characterized by the average dielectric parameters of normal breast tissue. After the antenna is excited with the UWB pulse, field amplitudes are computed at points along a line perpendicular to the antenna and passing through the feed. The results are linearly interpolated to provide estimates of the total spreading and loss at the desired distances from the antenna.

The reconstructed image is created by time-shifting and summing data points from the $M$ calibrated, integrated, and compensated waveforms $C_1, C_2, \ldots, C_M$ for each synthetic focal point in the breast. First, distances from each antenna to the focal point are computed and converted into time delays. The time delays are used to identify the contribution from each processed signal. All contributions are summed and the squared value of this sum is assigned to the pixel value at the focal point:

$$I(\vec{r}) = \left[ \sum_{m=1}^{M} C_m(\tau_m(\vec{r})) \right]^2$$

(2.1)

where $C_m$ is the post-processed backscatter waveform at the $m^{th}$ antenna located at $\vec{r}_m$, and $\tau_m(\vec{r}) = 2 | \vec{r} - \vec{r}_m | / (v \Delta t)$ is the discrete time delay from the $m^{th}$ antenna to the synthetic focal point at $\vec{r}$. Here, $v$ is the assumed velocity of propagation of the signal in the medium, calculated by assuming that the breast tissue is homogeneous with the average permittivity of normal breast tissue. The focal point is scanned to a new location in the region of interest, and this process is repeated. Each value of $I$ is converted to a pixel using an appropriate colormap.
scheme.

In a 3-D system with a realistic antenna, further improvement can be made to the algorithm by compensating for the non-omnidirectional antenna radiation pattern. The implementation and evaluation of this approach might be of future research interest.

2.2 Imaging results using simple 3-D numerical breast models with resistively loaded bowtie antennas

The simple beamforming algorithm presented in Section 2.1 is first tested with simulated backscattered waveforms generated from a simplified 3-D numerical breast phantom, where the breast is modeled as a half-space of heterogeneous breast tissue bounded by a 2-mm-thick layer of skin (Fig. 4). The dielectric properties of normal breast tissue are assigned random variations of up to ±10% around the nominal values of $\epsilon_r = 9$ and $\sigma = 0.4$ S/m, distributed over 4-mm cubes. The dielectric properties for a spherical 6-mm-diameter tumor introduced 3.3 cm below the skin surface are $\epsilon_r = 50$ and $\sigma = 4$ S/m. Thus, the assumed contrast between malignant and normal breast tissue is approximately 5:1 in relative permittivity and 10:1 in conductivity. The ±10% variation represents the expected variation in real breast tissue. The skin is assigned the following values: $\epsilon_r = 36$, $\sigma = 4$ S/m. The frequency dependence of the dielectric properties of all tissue types is neglected in our simplified 3-D models. The antenna is backed by an impedance matching layer of lossy liquid with the average dielectric-properties parameters of
normal breast tissue.

Figure 4: Model geometry for the simplified 3-D numerical breast phantom (side view).

The ultrawideband antenna element is a 2-cm-long resistively loaded bow-tie (a smaller version of that presented in [3]). The bow-tie antenna is placed directly on the skin and moved to 41 locations to create a synthetic planar array in the simulations. The array is arranged in five rows of five positions each interleaved with four rows of four positions each (Fig. 5). The array spans 6 cm in the x-direction (defined as the distance between the antenna feed locations marked by the small circles in Fig. 5) and 8.2 cm in the y-direction. At each location, the antenna is excited with a differentiated Gaussian pulse with full-width half-maximum (FWHM) of 170 ps in time and approximately centering at 4 GHz in frequency with a 6GHz bandwidth. A 50-Ω resistive voltage source is modeled at the feed point. During and following excitation, the current at the antenna feed is recorded in the simulation.
Step 1 and 2 described in Section 2.1 are applied to the processed data, and the resulting images are shown for the three orthogonal planes defined in Fig. 6. Reconstructed images of tumor-bearing breast models are presented in Fig. 7. The tumor is easily detected in the reconstructions, illustrating the enhancement of the tumor response via coherent addition of returns from the scattering object (the malignant tumor). Images of tumor-free breast models show no evidence of strong scatterers, confirming that the returns from spatially distributed heterogeneities in normal breast tissue are added incoherently. The signal-to-clutter ratio ($S/C$), defined as the ratio of the maximum tumor response to the pixel intensity at the same location in the image of a tumor-free model, is found to be about 21 dB in the 3-D reconstructed image. This demonstrates the significant difference in images obtained for the tumor-free and tumor-bearing models. The location of the maximum tumor response is $(x = 75 \text{ mm}, y = 75 \text{ mm}, z = 59 \text{ mm})$, which closely
Figure 6: Definition of the three orthogonal reconstruction planes.

resembles the physical location of the center of the tumor \((x = 75\, \text{mm}, y = 75\, \text{mm}, z = 58\, \text{mm})\). The FWHM response, a measure of the physical extent of the tumor, is found to be 8.0 mm in the x-direction, 6.5 mm in the y-direction, and 4.5 mm in the z-direction. This FWHM response is only 10% larger in volume than the 6-mm-diameter spherical tumor placed in the model. Additional details, including a comparison between the planar and cylindrical configuration, can be found in [2].
Figure 7: 3-D reconstructed image shown in three orthogonal planes: (a) axial, (b) sagittal, (c) coronal. The linear scale for (a)-(c) is shown below c).
2.3 Imaging results using anatomically realistic 2-D numerical breast models

Next, the simple beamforming algorithm is tested with simulated backscatter waveforms generated from a 2-D anatomically realistic FDTD model of the naturally flattened breast [1]. The FDTD model is derived from high-resolution MRI data sets. Fig. 8 shows a sagittal cut through a high-resolution breast MRI data set obtained during a scan of a patient lying in a prone position. A low-resolution scan was taken with the same patient in a supine position. Using the low-resolution face-up MRI data set as a guide, we vertically compressed and laterally expanded the high-resolution face-down images so that the overall shape of the breast matched that of the naturally flattened breast in the face-up position. We segmented the skin layer and removed the subtle MRI artifacts adjacent to the skin. Finally, we used a linear interpolation scheme to change the MRI pixel size ($0.625 \times 0.625 \text{ mm}^2$) to the desired FDTD grid cell size ($0.5 \times 0.5 \text{ mm}^2$). The resulting FDTD grid is terminated with PML absorbing boundary conditions (see Ch. 7 in [4]).

The frequency dependence of $\epsilon_r$ and $\sigma$ has been incorporated into the FDTD simulations using an auxiliary differential equation approach (see Ch. 9 in [4]). The Debye parameters of Eq. 1.1 have been chosen to fit the published data as presented in Fig. 1 over the band of interest (100 MHz to 20 GHz). Fig. 9 shows the spatial variation of the dielectric properties at 6 GHz. To preserve the heterogeneity of normal breast tissue, we linearly mapped the range of MRI pixel densities in the breast interior to a range of Debye parameters representing a variation of $\pm 10\%$ around an estimated baseline ($\epsilon_{r,\text{avg}} = 9.8$, $\sigma_{\text{avg}} = 0.4 \text{ S/m}$ at 6 GHz). This variation represents an upper bound on previously reported breast tissue...
variability ([5], [6]), which presumably covers the range from fat to fibroglandular tissue. Thus, regions of dense fibroglandular tissue (the darkest pixels in Fig. 8) were assigned Debye parameters yielding the largest values of $\epsilon_r$ and $\sigma$. Similarly, regions of fatty tissue (the lightest pixels in Fig. 8) were assigned the smallest values. Fig. 10 shows the assumed frequency dependence of the dielectric constant and conductivity of breast tissue in our baseline FDTD model. The thin solid curve represents $\varepsilon_r(\omega)$ and $\sigma(\omega)$ for the malignant lesion. The thick shaded curve illustrates the assumed $\pm 10\%$ variation around the average dielectric properties of normal breast tissue. The grayscale shading of this curve corresponds to that used for normal breast tissue in Fig 9. The Debye model for the average complex permittivity of normal breast tissue yields $\varepsilon_r = 9.8$ and $\sigma = 0.4$ S/m at 6 GHz, the spectral peak of the ultrawideband input pulse. For malignant breast tissue, the Debye model yields $\varepsilon_r = 50.7$ and $\sigma = 4.8$ S/m at 6 GHz.
Figure 9: 2-D MRI-derived FDTD breast model containing a 2-mm-diameter malignant lesion at a depth of 3.1 cm. The 17 black dots along the surface of the breast represent antenna locations.

In the model shown in Fig. 9, the 2-mm-diameter malignant tumor has been artificially introduced at a depth of 3.1 cm below the surface of the 2-mm-thick skin layer (with $\varepsilon_r = 36$ and $\sigma = 4$ S/m). A conformal antenna array consisting of 17 elements modeled as hard sources is located on the surface of the breast along the span-axis between 1.0 cm and 9.0 cm. The location of each antenna is marked by a black dot in Fig. 9. The antenna array is backed with a synthetic material matching the average dielectric properties of normal breast tissue at 6 GHz.

In the FDTD simulation, the simulated scan involves exciting each antenna individually with a 110-ps differentiated Gaussian pulse and recording the backscatter current response at the same antenna element.

The coherent-addition process described by Eq. 2.1 is illustrated in Fig. 11, which shows a subset of calibrated, integrated backscatter waveforms computed from an FDTD model containing a 6-mm-diameter malignant tumor at a depth of
Figure 10: Modeled permittivity (left panel) and conductivity (right panel) as a function of frequency for malignant (thin solid curve) and normal (thick shaded curve) breast tissue.

3.3 cm. Two sets of time delays are marked on the backscatter waveforms. The set of asterisks show the time delays estimated for a synthetic focal point located within the region of the tumor. These marked field values will add coherently. The other set of time delays corresponds to a synthetic focal point located 3 cm to the right of the tumor, at a depth of 1.3 cm. Those marked field values will add incoherently.

The synthetic focal point is scanned throughout the breast in increments of 1 mm². Each value of $I$ is converted to a pixel intensity and displayed using an appropriate mapping function. The reconstructed microwave image is shown in Fig. 12. The bright spot located at a depth of approximately 3 cm is at precisely the location of the 2-mm-diameter malignant tumor present in the FDTD model. The image also shows the suppression of clutter signals generated by the tissue heterogeneity in the surrounding regions. We have estimated the signal-to-clutter ($S/C$) ratio, again defined as the ratio of the maximum tumor response to the
pixel intensity at the same location in the image formed without a tumor present, to be approximately 11 dB for the breast model containing the 2-mm-diameter tumor. The minimum signal-to-clutter ratio (minimum $S/C$), defined as the ratio of the maximum tumor response to maximum clutter response appearing in the reconstruction of tumor-free model, is 8 dB. We also calculated the lateral full-width of the tumor response at the half-maximum to be on the order of 5 mm.

Our FDTD simulations have demonstrated that computationally efficient signal processing techniques can be applied to microwave backscatter data in a straightforward manner that avoids solving the inverse scattering problem.
Figure 11: Processed backscatter waveforms computed for an MRI-derived breast model containing a 6-mm-diameter tumor at a depth of 3.3 cm. The asterisks (*) mark the time delays corresponding to a synthetic focal point located within the tumor; the triangles (△) mark the time delays corresponding to a focal point away from the tumor.
Figure 12: 2-D microwave breast image reconstructed from the processed backscatter waveforms computed for the numerical breast phantom shown in Fig. 9.
Bibliography


Chapter 3

Advanced Microwave Imaging via Space-Time (MIST) Beamforming

In Chapter 2, we demonstrated the efficacy of spatially focusing the backscattered signals. However, the simple delay-and-sum beamforming approach lacks the capacity to compensate for frequency-dependent propagation effects and has limited ability to discriminate against artifacts and noise. In this chapter, we introduce an improved MIST beamforming approach which spatially focuses the backscatter waveforms to discriminate against clutters generated by the heterogeneity of normal breast tissue and noise while compensating for frequency-dependent propagation effects. An improved data-adaptive algorithm for removing artifacts in the received signals is also presented. The performance and robustness of the MIST beamforming algorithm is evaluated using anatomically realistic numerical breast phantoms.

3.1 Artifact removal algorithm

In this section, we present an algorithm for optimally removing artifacts in the received signals due to incident pulse, antenna reverberation, and backscatter from the skin-breast interface. Consider an array of \( N \) antennas and denote the received
signal at the \(i^{th}\) antenna as \(b_i(t)\). Each received signal is converted to a sampled waveform, \(b_i[n]\). The artifact components in the \(N\) channels are similar but not identical due to local variations in antenna performance, skin thickness, and breast heterogeneity. As an improvement over the approach of subtracting the averaged waveform across the \(N\) channels from each channel, as presented in Section 2.1, we can compensate for channel-to-channel variation by estimating the artifact in each channel as a filtered combination of the signal in all other channels. The filter weights are chosen to minimize the residual signal mean-squared error over that portion of the received data dominated by the early-time artifact response [1].

![Block diagram illustrating the algorithm for removing the artifact from the backscattered signal received at the first of \(N\) antennas.](image)

Without loss of generality, suppose that the artifact is to be removed from the first antenna. Figure 13 illustrates the artifact removal procedure. The artifact response at the \(n^{th}\) sample in the first channel is estimated from \(2J + 1\) successive samples centered on the \(n^{th}\) sample in each of the other \(N - 1\) channels. The number of samples \((2J + 1)\) is determined empirically. Define the \((2J + 1) \times 1\)
vector of time samples in the $i^{th}$ antenna channel as

$$b_i[n] = \begin{bmatrix} b_i[n-J], \ldots, b_i[n], \ldots, b_i[n+J] \end{bmatrix}^T, \quad 2 \leq i \leq N \quad (3.1)$$

and let $b_{2N}[n] = \begin{bmatrix} b_2^T[n], \ldots, b_N^T[n] \end{bmatrix}^T$ be the concatenation of data in channels 2 through $N$. Similarly, let $q_i$ be the $(2J+1) \times 1$ vector of finite-impulse response (FIR) filter coefficients in the $i^{th}$ channel and $q = [q_2^T, \ldots, q_N^T]^T$ be the concatenation of FIR filter coefficients from channels 2 through $N$. The optimal filter weight vector $q$ is chosen to satisfy

$$q = \arg \min_{q} \sum_{n=n_0}^{n_0+m-1} \left| b_1[n] - q^T b_{2N}[n] \right|^2$$

where the time interval $n = n_0$ to $n = n_0 + m - 1$ represents the initial portion of the data record containing artifact and no backscattered signals from lesions. The solution to this minimization problem [2] is given by

$$q = R^{-1}p \quad (3.3)$$

$$R = \frac{1}{m} \sum_{n=n_0}^{n_0+m-1} b_{2N}[n]b_{2N}^T[n] \quad (3.4)$$

$$p = \frac{1}{m} \sum_{n=n_0}^{n_0+m-1} b_{2N}[n]b_1[n] \quad (3.5)$$

The fact that there is a high degree of similarity among the artifact components in all $N$ channels results in the sample covariance matrix $R$ being ill-conditioned. If $R$ is ill-conditioned, then the matrix inversion in (3.3) can result in a solution for $q$ that has very large norm and thus, amplifies noise. In order to prevent this, we replace $R$ with the low rank approximation

$$R_p = \sum_{i=1}^{p} \lambda_i u_i u_i^T \quad (3.6)$$
where $\lambda_i, 1 \leq i \leq p$, are the $p$ significant eigenvalues and $u_i, 1 \leq i \leq p$, are the corresponding eigenvectors. The filter weight vector is determined by replacing $R^{-1}$ in (3.3) with

$$R_p^{-1} = \sum_{i=1}^{p} \frac{1}{\lambda_i} u_i u_i^T$$  

(3.7)

The artifact is then removed from the entire data record of the first channel to create artifact-free data $x_1[n]$ given by

$$x_1[n] = b_1[n] - q^T b_{2N}[n]$$  

(3.8)

This algorithm introduces a small level of distortion in the backscattered signal from the lesion because the tumor response in the other $N - 1$ channels is added back in to the first channel. This is explicitly shown by decomposing $b_1[n]$ and $b_{2N}[n]$ into artifacts $s_1[n]$ and $s_{2N}[n]$ and residuals $d_1[n]$ and $d_{2N}[n]$, respectively. The values $n_0$ and $m$ are chosen so that $q$ is determined from a portion of the data in which the residuals are negligible and thus,

$$s_1[n] - q^T s_{2N}[n] \approx 0$$  

(3.9)

However, expanding $b_1[n]$ and $b_{2N}[n]$ in (3.8) in terms of artifacts and residuals gives

$$x_1[n] = s_1[n] - q^T s_{2N}[n] + d_1[n] - q^T d_{2N}[n]$$  

(3.10)

$$\approx d_1[n] - q^T d_{2N}[n]$$  

(3.11)

Thus the residual signal is distorted by $q^T d_{2N}[n]$. This term is generally small because $q$ tends to “average” across the $N$ channels and the signals in $d_{2N}[n]$ that
represent backscatter from the lesion are not aligned in time, thus they do not add in phase. If the residual $d_{2N}[n]$ were available, one could remove the distortion by filtering $d_{2N}[n]$ with $q$ and adding to $x_1[n]$. We may approximate this correction step by using $x_{2N}[n]$ to approximate $d_{2N}[n]$ where

$$
x_{2N}[n] = \left[ x_2[n-J], \ldots, x_2[n+J], \ldots, x_N[n-J], \ldots, x_N[n+J] \right]^T \tag{3.12}
$$

is the vector containing the data from the other $N-1$ channels after the artifact has been removed from each of them. That is, we reduce the distortion in $x_1[n]$ by replacing $x_1[n]$ with $\tilde{x}_1[n]$ where

$$
\tilde{x}_1[n] = x_1[n] + q^T x_{2N}[n] \tag{3.13}
$$

The performance of the artifact removal algorithm is illustrated by applying the algorithms to backscatter data obtained from anatomically realistic numerical breast phantom similar as described in Section 2.3. The 17 simulated backscatter waveforms are down-sampled from a sampling frequency of 1200 GHz to 50 GHz. All 17 received signals, both before and after removing the early-time artifact, are plotted in the left (early-time) panel of Fig. 14. A subset of the received signals ($b_i[n]$ and $\tilde{x}_i[n]$ where $i = 1, 3, 5, \ldots, 15, 17$) are plotted in the right (late-time) panel. Prior to applying the artifact removal algorithm, the early-time response, shown by the solid curves in the left panel, is dominated by the incident pulse and backscatter at the skin-breast surface (since the waveforms are obtained from a 2-D simulation, the antenna reverberation is not an issue). The late-time response, shown by the solid curves in the right panel using an enlarged vertical

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1. The differences are twofold. First, the antenna array is backed with a material with similar dielectric properties as skin instead of normal breast tissue. Second, current sources are used instead of hard sources.
scale, contains the tumor response and clutter due to heterogeneity in the breast. The incident pulse and skin backscatter response together are three orders of magnitude larger than the strongest tumor response. The left panel shows that the early-time response in the 17 channels are very similar but not identical. The dashed curves represent the processed signals, \( \tilde{x}_i[n] \), after applying the artifact removal algorithm. The results demonstrate that the early-time response is almost completely eliminated while the late-time tumor response is preserved in each channel. The removal of antenna response and skin artifact using the same algorithm is demonstrated in Chapter 5 with experimental results.
Figure 14: FDTD computed backscattered signals before applying the artifact removal algorithm (solid curves) and after (dashed curves). The left panel shows the early-time response while the right panel shows the late-time response.
3.2 Space-time beamforming procedure

The image of backscattered energy as a function of scan location \( \mathbf{r} \) is obtained by applying a space-time beamformer to focus the backscattered signals at each scan location. Chapter 2 illustrates the potential of imaging millimeter-sized breast lesions using simple time-shift and summing techniques. However, this simple beamforming scheme does not compensate for frequency-dependent propagation effects and thus has limited capability to discriminate against artifacts and noise. This section introduces our improved space-time beamforming algorithm.

We consider the design of a space-time beamformer for a specific scan location \( \mathbf{r}_0 \). Our goal is to design the beamformer to pass backscattered signals from \( \mathbf{r}_0 \) with unit gain while attenuating signals from other locations [3]. Figure 15 shows the beamforming procedure.

First, the received signals are time-shifted to align the returns from a hypothesized scatterer at a candidate location. The pre-processed signal in the \( i^{th} \) channel \( \tilde{x}_i[n] \) is delayed by an integer number of samples \( n_i(\mathbf{r}_0) = n_a - \tau_i(\mathbf{r}_0) \) so that the signals in each channel are approximately aligned in time. Here \( \tau_i(\mathbf{r}_0) \) denotes the round-trip propagation delay for location \( \mathbf{r}_0 \) in the \( i^{th} \) channel, computed by dividing the round-trip path length by the average speed of propagation and rounding to the nearest sample; \( n_a \) is the reference time to which all received signals are aligned. We choose \( n_a \) as the worst case delay over all channels and locations, that is,

\[
n_a \geq \text{round}(\max_{i,\mathbf{r}_0} \tau_i(\mathbf{r}_0)) \quad (3.14)
\]

The time aligned signals are windowed before the filtering stage to remove
interference and clutter that is present prior to time \( n_a \) using the window function

\[
g[n] = \begin{cases} 
    1 & n \geq n_a \\
    0 & \text{otherwise}
\end{cases} 
\] (3.15)

Then a bank of finite-impulse response (FIR) filters are applied to the windowed waveforms \( x'_i[n] \), one in each antenna channel. The purpose of the FIR filters is to equalize path length dependent dispersion and attenuation, interpolate any fractional time delays remaining in the backscattered lesion responses after coarse time alignment, and bandpass filter the signal. The FIR filters can be designed and implemented in either time domain [1] or frequency domain [4].

The shaded blocks in Fig. 15 (a) represent the time-domain FIR filter implementation procedure. The time-domain FIR filter in the \( i^{th} \) channel has coefficients represented by the \( L \times 1 \) vector \( w_i \). The filter length, \( L \), is chosen empirically to balance performance and complexity. The FIR filter weights are designed by solving the penalized least squares problem, which involves solving the inversion of an \( NL \times NL \) matrix, where \( N \) is the number of locations in the imaging domain. Detailed discussion of the FIR filter design can be found in [1]. The time-gated waveforms \( x'_i[n] \) are filtered by \( w_i \) using convolution, one in each channel. The filtered outputs \( z_i[n] \) are then summed across channels to produce the beamformer output \( z[n] \).

Alternatively, the FIR filters can be designed and applied in the frequency domain. In this approach, the shaded blocks in Fig. 15 (a) are replaced with steps described in Fig. 15 (b). First the time-domain waveforms \( x'_i[n] \) transformed to the frequency domain and pointwise multiplied by the frequency-domain beamformer coefficients \( W_i(l) \), where \( W_i(l) \) is the beamformer weight in the \( i^{th} \) channel at DFT.
frequency index $\ell$. The sum of these weighted signals forms the beamformer output $Z[n]$. An inverse DFT transforms the beamformer output back to the time-domain response $z[n]$. The beamformer coefficients $W_i(l)$ are obtained also by solving the penalized least-square problem in the design stage. However, it does not require matrix inversion in the frequency domain to do so. For detailed algorithm for filter design, refer to [4].

The FIR filter output $z[n]$ is then time-gated by window $h[r_0, n]$. Finally the output energy is calculated by

$$p(r_0) = \sum_n |z[n]h[r_0, n]|^2$$  \hspace{1cm} (3.16)

The purpose of applying window $h[r_0, n]$ is to reduce clutter effects by ensuring that the output energy is calculated using only samples of $z[n]$ containing backscattered lesion energy. A natural choice for the window is

$$h[r_0, n] = \begin{cases} 1 & n_h \leq n \leq n_h + \ell_h \\
0 & \text{otherwise} \end{cases}$$  \hspace{1cm} (3.17)

if the main lobe of tumor backscatter response occupies time points $n_h$ through $n_h + \ell_h$ in $z[n]$. In practice, scattering from the tumor is frequency-dependent, so the backscattered signal is a distorted version of the transmitted pulse. These dispersive effects increase the duration of the backscattered signal and complicate window selection. Our preliminary investigations suggest that the extent of the increase in duration is directly proportional to the tumor size. Since we are interested in detecting very small lesions, we have chosen to design $h[r_0, n]$ assuming a point scatterer model. This gives the largest possible signal-to-clutter ratio (S/C) for small tumors. The S/C for larger tumors is reduced by this choice; however, the backscattered signal from larger tumors is much stronger so a compromised
S/C is relatively inconsequential for tumor detection. Tapered windows such as a raised cosine or decaying exponential could also be used to preserve signal energy while discriminating against clutter.

The reconstructed image of microwave scattering strength is obtained by scanning $r_0$ throughout the reconstruction region and plotting beamformer output energy as a function of location.
Figure 15: (a) Block diagram illustrating the MIST beamforming process for location $r_0$ in the breast with time-domain FIR filter implementation (shaded blocks). (b) Block diagram of frequency-domain FIR filter implementation.
3.3 2-D beamformer design and performance

In order to illustrate the MIST beamforming algorithm presented in Section 3.2, the design and performance of a 2-D space-time beamformer will be discussed in this section. In our example, the 2-D beamformer is designed for a 1-D conformal antenna array containing 17 elements spanning 8 cm horizontally along the surface of the breast. The 2-D breast region which the beamformer is designed to scan encompasses a span of 10 cm and a depth of 4 cm. The transmitted pulse is a differentiated Gaussian with a full width at half maximum equal to 110 ps. The spectrum of this pulse has a peak near 6 GHz and significant energy between 1 and 11 GHz.

In the time-domain beamforming approach, the beamformer is designed over the range of frequencies of 1 to 11 GHz assuming a 50 GHz sampling frequency. The length of each FIR filter is $L = 45$. The design location $r$ is scanned over the breast region using a grid resolution of 1 mm. The post-beamformer window described by Eq. (3.16) is six sampling intervals in length, spanning 120 ps.

Figure 16 illustrates the ideal spatial discrimination capability of this 2-D beamformer. The beamformer gain, defined as the output power due to an idealized point scatterer in a homogeneous medium, is plotted on a dB scale as a function of scatterer position for three different design locations. Although these patterns will deteriorate in the presence of noise and clutter, they are valuable for illustrating the target performance of the beamformer. Figure 16 (a) shows the gain pattern with a peak at $(5.0 \text{ cm}, 3.1 \text{ cm})$ for the case when the beamformer is designed to pass backscattered signals originating from that location with unit gain. In Figure 16 (b), the design location $(5.0 \text{ cm}, 1.1 \text{ cm})$ is 2.0 cm shallower than that of
Figure 16 (a). Figure 16 (c) shows the gain pattern for a design location (8.0 cm, 2.1 cm) that is off-center with respect to the antenna array. These patterns show that the beamformer attenuates scattered signals originating from any location that is greater than 1 cm away from the design location by more than 20 dB.

Figure 17 (a) depicts the scanned beamformer output energy for the breast model of Fig. 9 after applying an idealized artifact removal algorithm. This idealized algorithm simply subtracts channel-by-channel the exact artifact component, that is, the backscatter signal recorded during an FDTD simulation of a tumor-free homogeneous breast model. Obviously, this idealized approach for removing the early-time artifact cannot be used in practice. However, in these simulated tests, it serves as a useful benchmark of the best performance possible. The origin of the dominant energy in Fig. 17 (a), localized around (5.0 cm, 3.2 cm), is the dielectric-properties contrast between malignant and normal breast tissue. The origin of the low-level energy spatially distributed throughout the image is the heterogeneity of normal breast tissue in the numerical breast phantom. The tumor is clearly detectable as it stands 18 dB above the maximum clutter in the corresponding image for a tumor-free model (\( S/C = 18 \text{ dB} \)). We compare the performance of the MIST beamforming method with the simple time-shift-and-sum beamforming scheme presented in Ch. 2. After applying the idealized artifact removal algorithm, the simple beamforming algorithm is applied to the backscatter waveforms and the resulting image is plotted in Fig. 17 (b) using the same colormap scheme as Fig. 17 (a). The \( S/C \) is 9 dB in this case, compared to the 18 dB obtained using MIST beamforming. This significant improvement is a consequence of accounting for frequency-dependent propagation effects and improved discrimination against clutter.
Figure 18 (a) depicts the scanned beamformer output energy for the same scenario after the early-time artifact removal algorithm described in Section 3.1 is applied to the backscattered data. The tumor stands above the maximum background clutter by approximately 17 dB. Comparison of Figures 17 (a) and Fig. 18 (a) clearly shows that the skin artifact is removed in Fig. 18 (a) at the expense of a small amount of of energy from the tumor response spreading throughout the image in the vicinity of the tumor. Figures 18 (b) and 18 (c) repeat the scenario of Fig. 18 (a) for different tumor locations. The image in Fig. 18 (b) is based on the backscattered signals computed using a model similar to Fig. 9 with the 2-mm-diameter tumor located at a depth of only 1.1 cm. In the model associated with Fig. 18 (c), the tumor was located 3.0 cm off the center axis at a depth of 2.1 cm. For the case when the modeled tumor is centered at (5.0 cm, 1.1 cm), the S/C is 17 dB. The peak of the tumor response occurs at (5.0 cm, 1.3 cm). When the modeled tumor is centered at (8.0 cm, 2.1 cm), the S/C is 18 dB, and the peak of the tumor response occurs at (8.0 cm, 2.3 cm).

Figure 19 depicts the beamformer output energy for two adjacent 2-mm-diameter tumors separated by 1.5 cm with the deeper tumor located at a depth of 3.1 cm. Two distinct scattering objects are clearly evident. The tumor response closer to the skin has S/C of 18 dB while the tumor response farther from the skin has S/C of 16 dB. This example illustrates the resolving capability of MIST beamforming.

In each of the images of Figures 18 and 19, the peak of the tumor response occurs 2 to 3 mm deeper than the true center of the tumor. This small localization error is a consequence of assuming that the scatterer is essentially a point-scatterer
in the beamformer design and of an inherent bias due to the beamformer compensating for deeper locations.

The performance of 2-D frequency-domain beamformer has similar performance, as presented in [4]. The 3-D frequency-domain beamformer performance is illustrated in Section 5-5.2.
Figure 16: Beamformer gain as a function of position in a 10 cm × 4 cm plane of the breast for the following design locations: (a) (5.0 cm, 3.1 cm), (b) (5.0 cm, 1.1 cm), (c) (8.0 cm, 2.1 cm). The first and second coordinates in each pair represent span and depth, respectively. In each pattern, the location of the maximum is equal to the design location and is marked by a ‘+’.
Figure 17: Color images showing the backscattered energy on a dB scale for numerical breast phantoms similar to Fig. 9 with a 2-mm-diameter malignant tumor centered at (5.0 cm, 3.1 cm). The artifact was removed with an idealized algorithm. (a) Image created by the MIST beamforming scheme. (b) Image created by using the simple time-shift-and-sum beamforming scheme.
Figure 18: Color images showing the backscattered energy on a dB scale for numerical breast phantoms similar to Fig. 9 with a 2-mm-diameter malignant tumor centered at (a) (5.0 cm, 3.1 cm), (b) (5.0 cm, 1.1 cm), and (c) (8.0 cm, 2.1 cm). The artifact removal algorithm of Section 3.1 and the MIST beamforming scheme of Section 3.2 were applied.
3.4 Robustness of MIST beamforming

As discussed in [1], the beamformers are designed assuming simple propagation models for a homogeneous breast medium with frequency-independent average dielectric properties. However, the actual backscatter data is acquired either via realistic FDTD simulations as done here or via physical measurements as would be done in an actual patient scan. In either case, the breast tissue is heterogeneous and its dielectric properties are frequency-dependent. These represent significant deviations from the simple propagation physics assumed in the beamformer design. Thus, the successful detection of small lesions in the test cases presented here demonstrates that our MIST beamforming method is inherently robust to deviation between actual propagation effects and assumed propagation models.
Several additional issues related to robustness need to be addressed. In practice, the average density of normal breast tissue and the degree of heterogeneity will vary from patient to patient within a certain margin. Therefore, we investigate the robustness of our MIST beamforming approach with respect to the hypothetical variations in the characteristics of normal breast tissue. We consider scenarios where the average dielectric properties of the normal breast are greater than that suggested by data in the literature as well as scenarios where the variability about the average dielectric properties is greater than that suggested in the literature. Tumor detection under these scenarios is inherently more challenging because of greater signal attenuation, increased clutter, and reduced contrast between malignant and normal tissue.

Table 1 presents 25 different numerical breast phantoms used throughout this investigation. A 2-mm-diameter tumor is located at (5.0 cm, 2.1 cm) in each phantom. The dispersive properties of normal tissue are incorporated in each numerical breast phantom using Eq. 1.1 and an appropriate set of Debye parameters; however, for ease in presentation of the data, Table 1 displays single-frequency dielectric property values calculated at the spectral peak of the input pulse. The cell in the upper left corner of the table describes the normal-breast-tissue characteristics of the baseline numerical breast phantom employed in Section 3.2.

Each column in Table 1 represents one of five different scenarios of normal breast tissue density. The average dielectric properties for the five scenarios are selected as follows. Starting with the baseline normal-tissue and malignant-tissue sets of Debye parameters, we identify six intermediate sets of Debye parameters that are uniformly spaced between the two extremes. Since the malignant tissue properties remain constant in these investigations, the contrast in the dielectric
constant at 6 GHz between malignant and average normal breast tissue decreases from 5.2:1 for the baseline scenario to 3.2:1, 2.4:1, 1.9:1, 1.5:1, 1.3:1, and 1.1:1 for the six intermediate scenarios. The smallest malignant-to-normal contrast in dielectric constant suggested in the literature is approximately 2:1 [5]. Therefore, the first four scenarios are sufficient to span the hypothesized range of normal breast tissue densities. For completeness, we study the first five scenarios.

Each row in Table 1 represents one of five different scenarios of breast tissue heterogeneity. The upper bound on the variation around the average is increased from ±10% to ±50%. The increased variability further diminishes the contrast between malignant and normal tissue. For example, in the breast phantom described by the last entry in the first column, the contrast between malignant and normal breast tissue decreases from 5.2:1 to 3.4:1 if the tumor is present in the densest region of the breast.

<table>
<thead>
<tr>
<th>Variability</th>
<th>$\varepsilon_{r_{av}} = 9.8$</th>
<th>$\sigma_{av} = 0.4$ S/m</th>
<th>$\varepsilon_{r_{av}} = 15.7$</th>
<th>$\sigma_{av} = 1.0$ S/m</th>
<th>$\varepsilon_{r_{av}} = 21.5$</th>
<th>$\sigma_{av} = 1.7$ S/m</th>
<th>$\varepsilon_{r_{av}} = 27.3$</th>
<th>$\sigma_{av} = 2.3$ S/m</th>
<th>$\varepsilon_{r_{av}} = 33.2$</th>
<th>$\sigma_{av} = 2.9$ S/m</th>
</tr>
</thead>
<tbody>
<tr>
<td>±10%</td>
<td>8.82 &lt; $\varepsilon_r$ &lt; 10.8</td>
<td>0.36 &lt; $\sigma$ &lt; 0.44</td>
<td>14.1 &lt; $\varepsilon_r$ &lt; 17.2</td>
<td>0.90 &lt; $\sigma$ &lt; 1.10</td>
<td>19.4 &lt; $\varepsilon_r$ &lt; 23.7</td>
<td>1.53 &lt; $\sigma$ &lt; 1.87</td>
<td>24.6 &lt; $\varepsilon_r$ &lt; 30.6</td>
<td>2.07 &lt; $\sigma$ &lt; 2.53</td>
<td>29.9 &lt; $\varepsilon_r$ &lt; 36.5</td>
<td>2.61 &lt; $\sigma$ &lt; 3.19</td>
</tr>
<tr>
<td>±20%</td>
<td>7.84 &lt; $\varepsilon_r$ &lt; 11.8</td>
<td>0.32 &lt; $\sigma$ &lt; 0.48</td>
<td>12.5 &lt; $\varepsilon_r$ &lt; 18.8</td>
<td>0.80 &lt; $\sigma$ &lt; 1.20</td>
<td>17.2 &lt; $\varepsilon_r$ &lt; 25.8</td>
<td>1.36 &lt; $\sigma$ &lt; 2.06</td>
<td>21.8 &lt; $\varepsilon_r$ &lt; 32.8</td>
<td>1.84 &lt; $\sigma$ &lt; 2.76</td>
<td>26.6 &lt; $\varepsilon_r$ &lt; 39.8</td>
<td>2.32 &lt; $\sigma$ &lt; 3.48</td>
</tr>
<tr>
<td>±30%</td>
<td>6.86 &lt; $\varepsilon_r$ &lt; 12.8</td>
<td>0.28 &lt; $\sigma$ &lt; 0.52</td>
<td>10.9 &lt; $\varepsilon_r$ &lt; 20.4</td>
<td>0.70 &lt; $\sigma$ &lt; 1.30</td>
<td>15.1 &lt; $\varepsilon_r$ &lt; 27.9</td>
<td>1.19 &lt; $\sigma$ &lt; 2.21</td>
<td>19.1 &lt; $\varepsilon_r$ &lt; 35.5</td>
<td>1.61 &lt; $\sigma$ &lt; 2.99</td>
<td>23.2 &lt; $\varepsilon_r$ &lt; 43.2</td>
<td>2.03 &lt; $\sigma$ &lt; 3.77</td>
</tr>
<tr>
<td>±40%</td>
<td>5.88 &lt; $\varepsilon_r$ &lt; 13.7</td>
<td>0.24 &lt; $\sigma$ &lt; 0.56</td>
<td>9.39 &lt; $\varepsilon_r$ &lt; 21.9</td>
<td>0.60 &lt; $\sigma$ &lt; 1.40</td>
<td>12.9 &lt; $\varepsilon_r$ &lt; 30.1</td>
<td>1.02 &lt; $\sigma$ &lt; 2.38</td>
<td>16.4 &lt; $\varepsilon_r$ &lt; 38.2</td>
<td>1.38 &lt; $\sigma$ &lt; 3.22</td>
<td>19.9 &lt; $\varepsilon_r$ &lt; 46.5</td>
<td>1.74 &lt; $\sigma$ &lt; 4.06</td>
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<tr>
<td>±50%</td>
<td>4.90 &lt; $\varepsilon_r$ &lt; 14.7</td>
<td>0.20 &lt; $\sigma$ &lt; 0.60</td>
<td>7.83 &lt; $\varepsilon_r$ &lt; 23.5</td>
<td>0.50 &lt; $\sigma$ &lt; 1.50</td>
<td>10.8 &lt; $\varepsilon_r$ &lt; 32.3</td>
<td>0.85 &lt; $\sigma$ &lt; 2.55</td>
<td>13.7 &lt; $\varepsilon_r$ &lt; 41.0</td>
<td>1.15 &lt; $\sigma$ &lt; 3.45</td>
<td>16.6 &lt; $\varepsilon_r$ &lt; 49.8</td>
<td>1.45 &lt; $\sigma$ &lt; 4.35</td>
</tr>
</tbody>
</table>

Table 1: Range of dielectric constant and conductivity values at 6 GHz for heterogeneous normal breast tissue in 25 different numerical breast phantoms.

First, we investigate the case where patient-specific (phantom-specific) estimates of the nominal dielectric properties of normal breast tissue are available and are incorporated into the beamformer design process. In this case, the average
Average dielectric properties at 6 GHz

\[
\begin{align*}
\varepsilon_{\text{avg}} &= 9.8 & \sigma_{\text{avg}} &= 0.4 \text{ S/m} \\
\varepsilon_{\text{avg}} &= 15.7 & \sigma_{\text{avg}} &= 1.0 \text{ S/m} \\
\varepsilon_{\text{avg}} &= 21.5 & \sigma_{\text{avg}} &= 1.7 \text{ S/m} \\
\varepsilon_{\text{avg}} &= 27.3 & \sigma_{\text{avg}} &= 2.3 \text{ S/m} \\
\varepsilon_{\text{avg}} &= 33.2 & \sigma_{\text{avg}} &= 2.9 \text{ S/m}
\end{align*}
\]

<table>
<thead>
<tr>
<th>Variability</th>
<th>\varepsilon_{\text{avg}}</th>
<th>\sigma_{\text{avg}}</th>
<th>\text{S}/\text{cm}</th>
<th>\text{dB}</th>
</tr>
</thead>
<tbody>
<tr>
<td>±10%</td>
<td>18.5</td>
<td>14.6</td>
<td>10.9</td>
<td>4.89</td>
</tr>
<tr>
<td>±20%</td>
<td>14.9</td>
<td>10.7</td>
<td>5.8</td>
<td>1.2</td>
</tr>
<tr>
<td>±30%</td>
<td>11.1</td>
<td>7.9</td>
<td>3.1</td>
<td>N/A</td>
</tr>
<tr>
<td>±40%</td>
<td>9.1</td>
<td>5.9</td>
<td>1.9</td>
<td>N/A</td>
</tr>
<tr>
<td>±50%</td>
<td>7.6</td>
<td>4.4</td>
<td>N/A</td>
<td>N/A</td>
</tr>
</tbody>
</table>

Table 2: \( S/C \) as a function of average dielectric properties and variability of dispersive normal breast tissue.

<table>
<thead>
<tr>
<th>Assumed dielectric properties</th>
<th>Average dielectric properties at 6 GHz of numerical breast phantom</th>
</tr>
</thead>
<tbody>
<tr>
<td>\varepsilon_{\text{avg}} = 9.8 &amp; \sigma_{\text{avg}} = 0.4 \text{ S/m}</td>
<td>(5.0, 2.0)</td>
</tr>
<tr>
<td>\varepsilon_{\text{avg}} = 15.7 &amp; \sigma_{\text{avg}} = 1.0 \text{ S/m}</td>
<td>(5.0, 2.6)</td>
</tr>
<tr>
<td>\varepsilon_{\text{avg}} = 21.5 &amp; \sigma_{\text{avg}} = 1.7 \text{ S/m}</td>
<td>(5.0, 3.0)</td>
</tr>
<tr>
<td>\varepsilon_{\text{avg}} = 27.3 &amp; \sigma_{\text{avg}} = 2.3 \text{ S/m}</td>
<td>(5.1, 3.4)</td>
</tr>
<tr>
<td>\varepsilon_{\text{avg}} = 33.2 &amp; \sigma_{\text{avg}} = 2.9 \text{ S/m}</td>
<td>(5.1, 3.4)</td>
</tr>
</tbody>
</table>

Table 3: Location of the detected tumor (span in cm, depth in cm) as a function of the dielectric properties assumed in the beamformer design and the actual average dielectric properties of the numerical breast phantom.

dielectric properties assumed in the beamformer design stage are matched with the actual average dielectric properties of the numerical breast phantom. Table 2 shows the \( S/C \) obtained for the 25 numerical breast phantoms of Table 1. Note that “N/A” is used for cases when the peak tumor energy is comparable to the background clutter. The \( S/C \) decreases as the degree of heterogeneity and/or average dielectric properties are increased, as expected. However, the 2-mm-diameter lesion embedded in heterogeneous normal breast tissue is detectable over a wide range of phantoms, most of which greatly exceed that which is expected in practice. Thus, the results show that our MIST beamforming method is effective even
when the average dielectric properties of normal tissue and/or the variability are substantially greater than that reported in the literature.

Second, we investigate the case where patient-specific (phantom-specific) estimates of the nominal dielectric properties of normal breast tissue are not available. In this case, the average dielectric properties assumed in the beamformer design are not matched with the actual average dielectric properties of the numerical breast phantom. For this investigation, the variability is fixed at ±10%, which corresponds to the numerical breast phantoms from the first row of Table 1. The constant dielectric constant and conductivity values assumed when designing the beamformer are varied over the same five scenarios as the actual average values. In Table 3, we show the location of the peak energy in the resulting image for each combination of actual and assumed average dielectric properties. As expected, the table shows that the tumor is most accurately localized when the actual average dielectric properties are equal to those assumed in the beamformer design. Figures 20 (a) - (d) show the images of MIST beamformer output energy for the results in the first four columns of the 3rd row of Table 3. In all cases, the tumor is still clearly detectable. The primary effect of mismatch between actual and assumed dielectric properties is a location bias. This bias is consistent with the corresponding differences in average speed of propagation.
(a)

(b)

(c)
Figure 20: Color images of backscattered energy plotted on a dB scale for four numerical breast phantoms each with a 2-mm-diameter malignant tumor centered at (5.0 cm, 2.1 cm). The average dielectric properties of normal breast for the four phantoms are (a) $\varepsilon_{\text{avg}} = 9.8$, $\sigma_{\text{avg}} = 0.4$ S/m; (b) $\varepsilon_{\text{avg}} = 15.7$, $\sigma_{\text{avg}} = 1.0$ S/m; (c) $\varepsilon_{\text{avg}} = 21.5$, $\sigma_{\text{avg}} = 1.7$ S/m; and (d) $\varepsilon_{\text{avg}} = 27.3$, $\sigma_{\text{avg}} = 2.3$ S/m at 6 GHz. The beamformer is designed assuming $\varepsilon_{r} = 21.5$ and $\sigma = 1.7$ S/m at 6 GHz.
Bibliography


Chapter 4

Design and fabrication of an ultrawideband transmitting/receiving antenna element

Chapter 2 and Chapter 3 present the theoretical feasibility study of the ultrawideband microwave breast imaging system by using 2-D and 3-D numerical breast phantoms. The first challenge of implementing this system is the design and fabrication of high-performance transmitting/receiving antennas. In this chapter, we report the numerical analysis and experimental characterization of an ultrawideband ridged pyramidal horn antenna with curved launching plane for radiating short microwave pulses. Detailed 3-D finite-difference-time-domain (FDTD) simulations have been conducted to assist with the characterization of the antenna. FDTD results are compared with experimental data and are shown to be in good agreement. We demonstrate that the antenna exhibits a very low voltage standing wave ratio ($\leq 1.5$) over a wide frequency range from 1 GHz to 11 GHz and the radiated pulse resembles the derivative of the source waveform very well. The spatial distribution of radiated energy is characterized both in the time domain, using
transient field observations at various angles, as well as in the frequency domain, using single-frequency far-field radiation patterns. We conclude that this antenna offers high-fidelity transmission and reception of ultrashort microwave pulses with minimal distortion.

4.1 Background

Our ultrawideband microwave breast imaging system requires antenna elements capable of radiating and receiving short microwave pulses. While frequency-domain antenna characterization standards are well defined and accepted, these standards become insufficient when describing the time-domain transmitting/receiving characteristics of pulse-radiating antennas. A number of time-domain antenna characteristics have been defined and used in the literature [1]. Here we do not attempt to discuss all aspects of time-domain antenna characteristics. Instead, we will examine several antenna design requirements that are important for our UWB microwave imaging application. These requirements are listed as below:

- The antenna should have low reflection at the antenna feed or any point along the antenna throughout the operation frequency (1–11 GHz).
- The radiated and received microwave pulse should be a faithful reproduction of the derivative of the excitation waveform at any direction inside of the main-beam of the radiation.
- The antenna should be directional. The width of the radiation-pattern main-lobe should be moderate. The sidelobes should be minimum.
- The dimension of the antenna should be compact to fit on the breast.
Typical examples of wideband antennas used for pulse radiation include resistively loaded monopoles, dipoles, bow-tie antennas, and various forms of horn antennas [2], [3], [4], [5], [6]. There are various challenges adapting these antenna designs directly to the implementation of our UWB microwave breast imaging system including size limitation, bandwidth performance, radiation pattern limitation, fabrication difficulty, and requirement of UWB balun transitions.

It has been noted that the bandwidth of horn antennas can be increased significantly by adding metallic ridges to the waveguide and flared sections [7]. Numerical and experimental investigations of pyramidal horn antennas with double ridges have been reported previously [8]. Here we investigate a modified version of this antenna in which the waveguide section is eliminated and one of the two ridges is replaced by a curved metallic plane terminated by resistors. The generic form of this configuration has been proposed in [9]. To the best of our knowledge, however, a detailed numerical analysis and experimental characterization of this type of antenna has not been reported in the literature. In this chapter, we present a numerical and experimental study of a specific realization of this design, wherein the antenna is customized to cm-scale dimensions for operation in the microwave frequency range from 1 GHz to 11 GHz.

4.2 Description of the antenna geometry

Figure 21 illustrates the antenna geometry. The antenna consists of a pyramidal horn radiation cavity, a metallic ridge, and a curved metallic launching plane terminated with resistors. The pyramidal horn is connected to the outer conductor of the coaxial feed and serves as the ground plane, providing a current return path.
Figure 21: Geometry of the antenna. (a) End view. (b) Side view. (c) Photo of the fabricated antenna. The adjacent ruler is marked in units of centimeters.
Because of the coaxial feed, this ground-plane configuration eliminates the need for an UWB balun. The ground plane also confines the main beam of the radiation pattern to the opening of the horn, ensuring a compact radiation pattern.

The launching plane is a curved planar structure connected to the central conductor of the coaxial feed. As shown in Figs. 21 (a) and (b), the launching plane curves toward one of the side walls of the pyramidal horn and tapers toward the feed point. Termination resistors are attached between the end of the launching plane and the side wall of the pyramidal horn. Microwave energy is directed and launched by this curved plane into the surrounding medium. The termination resistors suppress reflections from the end of the launching plane. Figures 21 (a) and (b) also show the shape of the ridge which is attached to the side wall opposite the curved launching plane. The top surface of the ridge curves toward the antenna aperture.

The dimensions of the horn are chosen based on the physical size requirements and the operating frequency range associated with the desired application. The curvature and shape of the launching plane, the thickness and the contour of the curved side of the ridge, and the termination resistors are the main factors influencing the input impedance of the antenna. These parameters are chosen to minimize reflections at the feed point as well as at any point along the antenna. In order to match the 50-Ω input impedance of the feeding coaxial cable, two 100-Ω termination resistors are connected in parallel near opposite corners of the launching plane (see Figs. 21 (a) and (c)). Chip resistors are used to minimize inductance. Other geometrical parameters of the antenna are optimized using simulations and experimental measurements. In the finished design, the pyramidal horn has a depth of 13 mm with a 25 mm-by-20 mm aperture. The maximum
width of the launching plane is 12 mm, and the thickness of the ridge is 2 mm. The antenna is fabricated using brass sheets and connected to a coaxial cable via an SMA connector.

4.3 Numerical and experimental characterization in free-space

We have developed a 3-D FDTD model of the antenna to assist with the characterization and optimization of the antenna performance (Fig. 22 (a)). The antenna geometry is modeled using a uniform space lattice with a 0.5-mm grid resolution. The curved or flared metal surfaces of the antenna are modeled using a staircased PEC approximation. The base of the pyramidal horn is covered by a small PEC plane to provide a complete ground plane. A vertical gap of one grid cell exists between the feed point of the launching plane and the base of the horn. The antenna excitation is implemented using a 1.0-V, 50-Ω resistive voltage source across the gap [10]. Figure 22 (b) shows the geometry at the feed point of the ftdtd model. The 100-Ω termination resistors attached to the end of the launching plane are also incorporated into the FDTD model as lumped circuit elements [10]. The FDTD grid is terminated with a Berenger perfectly matched layer (PML) absorbing boundary condition [11].

Data for our experimental characterization of the antenna is acquired using an Agilent E8364A performance network analyzer (PNA). All time-domain “measured” waveforms discussed below have been generated synthetically using the PNA in a swept frequency mode. Measurements are conducted from 1 GHz to 11 GHz using 201 frequency samples. Frequency-domain data is scaled by the
Figure 22: (a) FDTD model of the UWB antenna geometry. (b) cross-sectional side-view with the launching plane on the right side). The solid lines depict Yee $\vec{E}$ components assigned with dielectric properties of metal. The region bounded by the dashed rectangle is zoomed in in (c) to illustrate the geometry of the antenna feed. (c) Cross-sectional side-view of the antenna feed.
The UWB performance of the antenna can be characterized in the frequency domain by the input reflection coefficient (which in this case is equal to $S_{11}$) or the voltage standing wave ratio (VSWR). In our experiments, $S_{11}$ is recorded and converted to VSWR. In our FDTD simulations, a source voltage waveform with spectral content covering the desired frequency range is excited at the feed point. Current and voltage waveforms $v(t)$ and $i(t)$ are recorded at the source location in the grid and converted to a frequency-domain input voltage ($V_{in}$) and current ($I_{in}$) using a forward DFT. Then, the input impedance, $Z_{in}$, is calculated as $V_{in}/I_{in}$ and $S_{11}$ is calculated as $(Z_{in} - Z_o)/(Z_{in} + Z_o)$, assuming $Z_o = 50\Omega$ for the characteristic impedance of the feedline. Finally, the simulated $S_{11}$ is converted to VSWR. As shown in Fig. 23, both the measured and simulated VSWRs are less than 1.5 over the entire frequency range of interest.

The UWB performance of the antenna can be characterized in the time domain.
Figure 24: (a) Source waveform applied to the input terminals of the transmitting antenna. (b) FDTD-computed and measured waveforms recorded at the receiving antenna located at a distance of 5 cm from the transmitter.
by the fidelity \( F \), which is a measure of how accurately the transmitted waveform reproduces the time derivative of the voltage applied to the antenna terminals, or, equivalently, how accurately the received voltage reproduces the transient field incident upon the antenna [1]. To investigate the antenna’s fidelity in transmission and reception of the UWB signal, \( s(t) \), shown in Fig. 24 (a), two replicas of the antenna shown in Fig. 21 are connected to the two ports of the PNA and aligned face-to-face with a 50-mm separation between the ends of the pyramidal horns. The forward transmission coefficient \( S_{21} \) is measured and converted to a frequency-domain received voltage, \( V_2 = S_{21}E(\omega)/2 \), where \( E(\omega) \) is the spectrum of the desired source waveform, \( s(t) = e^{-(t-4\tau)^2/\tau^2}\sin(2\pi f_0(t - 4\tau)) \) with \( f_0 = 6 \text{ GHz} \) and \( \tau = 63 \text{ ps} \). Then \( V_2 \) is transformed to the time-domain waveform \( v_2(t) \) using an inverse DFT. In the FDTD simulation, the two antennas are modeled using a configuration similar to the experimental set-up. The source waveform, \( s(t) \), is applied at the input terminals of the transmitting antenna and the voltage \( v_2(t) \) at the receiving antenna is recorded directly. As shown in Fig. 24 (b), there is excellent agreement between the simulated and measured versions of \( v_2(t) \).

The fidelity, defined as \( F = \max_{\tau} \int_{-\infty}^{\infty} \hat{r}(t-\tau)\hat{f}(t)dt \), corresponds to the maximum magnitude of the cross-correlation between the normalized observed response \( \hat{r}(t) \) and the ideal response \( \hat{f}(t) \). A fidelity of \( F = 1 \) indicates a perfect match between \( \hat{r}(t) \) and \( \hat{f}(t) \). Here, \( \hat{f}(t) \) is calculated as the normalized time-derivative of the source waveform, \( s(t) \), and \( \hat{r}(t) \) is calculated using the normalized versions of the simulated and measured waveforms plotted in Fig. 24 (b). This calculation yields a fidelity of approximately 0.96 in both the simulation and experiment.

To examine the fidelity as a function of angle off boresight, we use FDTD simulations to compute the transient fields radiated by the transmitting antenna at
Figure 25: Electric-field waveforms computed as a function of observation angle at a constant distance of 5 cm from the transmitting antenna in free space. (a) E-plane waveforms with the launching plane positioned on the right side. (b) H-plane waveforms.
Figure 26: Simulated and measured far-field radiation patterns at 6 GHz. (a) E-plane pattern with the launching plane positioned on the right side. (b) H-plane pattern.

Several observation points. Figure 25 (a) shows the radiated electric field waveforms observed at a distance of 50 mm from the antenna over an angular span of 90 degrees on either side of boresight in the E-plane. Here, the E-plane intersects the launching plane and ridge and divides the antenna geometry into two symmetric halves. Figure 25 (b) shows the radiated field in the H-plane, which passes between the launching plane and ridge. The fidelity values calculated for all the displayed waveforms are equal to or greater than 0.92. The waveforms in Fig. 25 also illustrate the antenna’s directivity.

Finally, we investigate the directivity in the far field at discrete frequencies. For the radiation pattern measurements, the transmitting and receiving antennas are separated by a distance of 300 mm. The transmitting antenna is rotated in $5^\circ$ increments in both the E and H planes. The magnitude of $S_{21}$, which is proportional to the intensity of the field radiated by the transmitter, is recorded at a given frequency as a function of rotation angle. In the FDTD simulation, a
sinusoidal waveform with the desired frequency is excited at the feed point and a frequency-domain near-field-to-far-field transformation is used to calculate the far field pattern. Figure 26 shows good agreement between the simulated and measured antenna radiation patterns at 6 GHz. Because of the asymmetry in the antenna geometry, the electric field in the E-plane is concentrated in the region between the launching plane and the opposing side wall before being launched to free space. Consequently, the E-plane radiation pattern exhibits a slight tilt (approximately 15°) towards the ridge. The H-plane radiation pattern is symmetric about boresight because of the symmetry exhibited by the antenna in this plane. Simulations and measurements conducted at other frequencies indicate that the directivity increases with frequency and that the far-field radiation patterns exhibit a compact main lobe and minimum sidelobes over the entire frequency range of interest.

4.4 Numerical and experimental characterization in low-loss emersion-medium

For our space-time microwave imaging for breast cancer detection, the antenna needs to be immersed in a non-free-space medium to ensure reasonable coupling of the UWB signals into the biological tissue. Therefore, we have performed additional experiments and simulations with this antenna immersed in a low-loss liquid immersion medium—soybean oil ($\varepsilon_r = 2.6$, $\sigma = 0.05$ S/m at 6 GHz).

As shown in Fig. 27, the magnitude of both measured and simulated VSWR with the antenna immersed in soybean oil falls below 1.7 and remains relatively constant over the frequency range of interest.
Figure 27: Simulated and measured VSWR of the antenna when it is immersed in soybean oil.

The transmission measurement and simulation are also done to characterize the pulse-radiation/receiving capability. Again, two replicas of the antenna are connected to the two ports of the PNA and aligned face-to-face immersed in soybean oil with a 50-mm separation between the ends of the pyramidal horns. The forward transmission coefficient ($S_{21}$) is measured and converted to the time-domain waveform. In the FDTD simulation, the two antennas are modeled using a configuration similar to the experimental set-up. As shown in Fig. 28 (b), there is excellent agreement between the simulated and measured versions of $v_2(t)$. The Fidelity calculation yields a F of approximately 0.91 and 0.94 for measured and simulated data respectively.

To examine spacial distribution of the radiated energy, we use FDTD simulations to compute the transient fields radiated by the transmitting antenna at several observation points. Figure 29 (a) shows the radiated electric field waveforms observed at a distance of 50 mm from the antenna over an angular span of 90
Figure 28: (a) Source waveform applied to the input terminals of the transmitting antenna. (b) FDTD-computed and measured waveforms recorded at the receiving antenna located at a distance of 5 cm from the transmitter. Both antennas are immersed in soybean oil.

degrees on either side of boresight in the E-plane. Here, the E-plane intersects the launching plane and ridge and divides the antenna geometry into two symmetric halves. Figure 29 (b) shows the radiated field in the H-plane, which passes between the launching plane and ridge.
Figure 29: Electric-field waveforms computed as a function of observation angle at a constant distance of 5 cm from the transmitting antenna when it is immersed in soybean oil. (a) E-plane waveforms with the launching plane positioned on the right side. (b) H-plane waveforms.
Bibliography


Chapter 5

Experimental and Numerical Investigation of Tumor Detection in Multilayer Breast Phantoms

While extensive experimental results have been obtained using a pre-clinical prototype of a microwave tomographic system [1], only very preliminary experimental studies have been reported to date using UWB radar techniques. In a recent experimental feasibility study [2], [3], simple time-shift-and-sum focusing schemes were used to detect a two-dimensional wood, copper, or water-filled object (representing a malignant tumor) inside an otherwise hollow PVC pipe (representing skin and normal breast) in free space. In this study, the pipe was illuminated by a large horn antenna or resistively loaded monopole antenna positioned at several points encircling the pipe. The study was designed to mimic the system configuration where the patient is lying in a prone position with antennas surrounding the breast.

In this chapter, we present an in-depth 3-D experimental and numerical study of the MIST beamforming approach using multilayer breast phantoms. We summarize and demonstrate the efficacy of the artifact removal and MIST beamforming algorithms that are applied to the backscatter signals received from representative
breast phantoms. In practice, the average density of normal breast tissue and the degree of heterogeneity will vary from patient to patient, thereby introducing variability in the contrast between malignant and normal tissue. Therefore, in Section 5.4, we investigate the strength of the tumor response and the image signal-to-clutter ratios as function of the dielectric contrast. This study provides insights about how tumor detectability and system dynamic range requirements vary over the range of expected dielectric contrast.

5.1 Experimental setup and multilayer breast phantom configuration

![Figure 30: Schematic showing a cross-sectional side view of the experimental setup.](image)

The experiment setup shown in Fig. 30 emulates a system configuration where a patient is lying in a supine position with a 2-D antenna array placed near the
surface of the naturally flattened breast. The breast phantom consists of a container filled with a liquid mimicking normal breast tissue, a small synthetic tumor suspended in the liquid, and a thin layer of material representing the skin layer covering the normal breast tissue simulant. A single UWB antenna is sequentially repositioned in the horizontal plane using a computer-controlled mechanical x-y scanner to synthesize a 2-D antenna array placed above the skin. The antenna is immersed in a matching medium to couple microwave energy into the breast more efficiently. Here, for simplicity, the liquid used for the normal breast tissue simulant is also used as the immersion medium.

The tissue simulants in the phantom should be chosen to achieve a reasonable match to the dielectric properties of the corresponding tissue types. Most importantly, the dielectric contrast between the tissue simulants should mimic the contrasts observed between different biological tissues at microwave frequencies. As discussed in Chapter 1, normal breast tissue exhibits a relatively low dielectric constant ($\varepsilon_r$) and conductivity ($\sigma$) at microwave frequencies. As shown in Fig. 1, Debye models were fit to measured data for normal and malignant breast tissue ([4], [5], and [6]) in order to permit data extrapolation at higher microwave frequencies. Those Debye models predict a contrast of approximately 5:1 in $\varepsilon_r$ and 12:1 in $\sigma$ between malignant and normal breast tissue at 6 GHz. This contrast was used as the baseline scenario in our previous numerical studies. However, the average dielectric properties of normal breast tissue are expected to vary from patient to patient due to differences in the ratio between fat and fibroglandular tissue. Increased average density or heterogeneity of breast tissue would result in a reduced contrast with malignant tissue. In fact, the dielectric-properties profiles derived from recent microwave tomography experiments suggest that the contrast between
normal and malignant breast tissue may be closer to 2:1 in the RF/microwave frequency range [1]. Therefore, scenarios where the dielectric contrast between normal and malignant breast tissue simulants is reduced from the baseline case should be considered since tumor detection in those situations is inherently more challenging.

In addition to dielectric properties, other factors such as availability, cost, toxicity and stability also have to be taken into consideration in choosing phantom materials. In the study presented in this dissertation, soybean oil is used as the normal breast tissue simulant because it is an inexpensive, non-toxic liquid with dielectric properties similar to very low-water-content fatty tissue. The dielectric properties of the oil (\(\varepsilon_r = 2.6\) and \(\sigma = 0.05\) S/m at 6 GHz), as measured using an open-ended coaxial probe technique [7], fall slightly below the expected range of the dielectric properties for fatty breast tissue. Therefore, we have chosen materials for the skin and tumor simulants that similarly underestimate the actual dielectric properties of those tissue types, so that the dielectric contrasts in the breast phantom are more representative of those for actual tissue.

Five malignant tissue simulants with varying \(\varepsilon_r\) and conductivity \(\sigma\) are developed using a diacetin-water solution with different concentrations. The resulting contrast in \(\varepsilon_r\) between malignant and normal tissue simulants ranges from 1.5:1 to 5.2:1. As discussed previously, this represents the range of contrasts expected in clinical scenarios. Figure 31 (a) plots \(\varepsilon_{r,\text{malignant}}/\varepsilon_{r,\text{normal}}\) as a function of water content in the malignant tissue simulants. As the diacetin solution is diluted with more water, the mixture exhibits an increased \(\varepsilon_r\) and \(\sigma\) and thus yields a higher dielectric contrast with the normal breast tissue simulant. Figure 31 (b) shows the measured dielectric constant of normal and malignant tissue simulants for the
entire frequency range of interest (1 GHz–11 GHz). The synthetic tumor is made by pouring the water-diacetin mixture into a 4-mm-diameter cylindrical container that has a height of 4 mm. The dielectric properties of the cylindrical container are similar to those of the soybean oil. A 0.1-mm-diameter nylon thread is used to suspend the synthetic tumor in the oil.

The skin layer in the phantom is created using a 1.5-mm-thick unclad FR4 glass epoxy PC board. According to the manufacturer, the dielectric properties of FR4 at 1 GHz are $\varepsilon_r = 4.34$ and $\tan\delta = 0.016$. Thus the dielectric constant of the skin simulant falls in between that of the normal and malignant tissue simulants, as desired.

During data collection, the UWB antenna is sequentially scanned in 1-cm increments to 49 different positions in a 6-cm × 6-cm array. The antenna element is positioned so that its aperture is 1 cm above the skin surface. The antenna is connected to an Agilent E8364A (10 MHz–50 GHz) performance network analyzer (PNA) to transmit and receive microwave signals. At each antenna location in the synthetic array, the PNA performs a frequency sweep from 1 to 11 GHz with 201 frequency samples and records the backscatter ($S_{11}$ parameter). The frequency-domain backscattered signals are scaled by the spectrum of the desired input pulse and transformed to the time domain using an inverse FFT algorithm. In the results presented in this chapter, the input pulse is a modulated Gaussian pulse given by:

$$s(t) = e^{-\left(\frac{t-4\tau}{\tau}\right)^2} \sin(2\pi f_0(t - 4\tau))$$  \hspace{1cm} (5.1)

where $f_0 = 6$ GHz and $\tau = 80$ ps. The spectrum of this source waveform has a peak near 6 GHz and a 1/e bandwidth of 8 GHz, which is sufficiently covered by the 1–11 GHz swept frequency range.
Figure 31: Contrast in $\varepsilon_r$ at 6 GHz between normal and malignant breast tissue simulants. The horizontal axis shows the percentage of water (by volume) present in the water-diacetin solution used for the malignant tissue simulants. (b) Measured $\varepsilon_r$ of the normal breast tissue simulant and the five different malignant breast tissue simulants as a function of frequency.
5.2 Signal processing and image formation procedures

After measuring backscatter signals ($S_{11}$) from the multilayer breast phantoms, time-domain backscatter waveforms are synthesized. Next, dominant early-time artifacts are removed from the waveforms before 3-D MIST beamforming is employed to create an image of backscatter energy as a function of position.

The early-time artifacts in the received waveforms include antenna reverberation and reflections from the skin-breast interface. The data-dependent algorithm reported in [8] is applied to remove these artifacts. In this algorithm, the artifact in the waveform received by a single antenna is estimated as a filtered combination of the waveforms received at all other antenna locations and removed from the received waveform. The filter weights are chosen to minimize the residual signal mean-squared error calculated over the artifact-dominated early-time response.

The efficacy of the artifact-removal algorithm is demonstrated using backscatter waveforms collected from the experimental breast phantom illustrated in Fig. 30. The 4-mm-diameter synthetic tumor made of simulant #5 ($\varepsilon_r,\text{malignant} = 5.2 \varepsilon_r,\text{normal}$) is placed 2.0 cm below the skin surface under the central antenna location. In Fig. 32, the signals received at the central row in the antenna array are plotted before and after artifact removal. Prior to applying the artifact removal algorithm, the early-time response, shown by the dashed curves in the left panel, is dominated by the antenna reverberation and skin-breast backscatter response. The late-time response, shown by the dashed curves in the right panel using an enlarged vertical scale, contains the tumor response which is completely masked by
Figure 32: Backscattered signals recorded for the experimental breast phantom with tumor simulant #5. The waveforms received at the central row of the synthetic array are plotted before applying the artifact removal algorithm (dashed curves) and after (solid curves). The left panel shows the early-time response while the right panel shows the late-time response. The shaded regions highlight the expected time window of the tumor response.

The slowly decaying artifact response. The solid curves represent the processed signals obtained by applying the artifact removal algorithm. The early-time artifact is almost completely eliminated as shown in the left panel. The tumor response is now clearly evident in the late-time response depicted in the right panel. The shaded areas highlight the time window in which the tumor response is expected, based on the known material properties and location of the tumor.

The image of backscattered energy as a function of scan location \( r \) is obtained by applying a space-time beamformer to focus the backscattered signals at each scan location. For a specific scan location \( r_0 \), the goal of the beamformer is to
pass backscattered signals from \( r_0 \) with unit gain while attenuating signals from other locations \([9]\). The pre-processed signal in the \( i^{th} \) antenna location is discretized assuming a 50 GHz sampling frequency and denoted by \( \tilde{x}_i[n] \). First, these signals are time-shifted to align the returns from a hypothesized scatterer at a candidate location. Each signal \( \tilde{x}_i[n] \) is delayed by an integer number of samples \( n_i(r_0) = n_a - \tau_i(r_0) \) so that these signals are approximately aligned in time. Here \( \tau_i(r_0) \) denotes the round-trip propagation delay for location \( r_0 \) in the \( i^{th} \) channel, computed by dividing the round-trip path length by the average speed of propagation and rounding to the nearest sample; \( n_a \) is the reference time to which all received signals are aligned. We choose \( n_a \) as the worst case delay over all channels and locations, that is,

\[
n_a \geq \text{round}(\max_{i, r_0} \tau_i(r_0)) \quad (5.2)
\]

The time aligned signals are windowed before the filtering stage to remove interference and clutter that is present prior to time \( n_a \) using the window function

\[
g[n] = \begin{cases} 
1 & n \geq n_a \\
0 & \text{otherwise}
\end{cases} \quad (5.3)
\]

Then a bank of finite-impulse response (FIR) filters are applied to the windowed waveforms, one in each antenna channel. The purpose of the FIR filters is to equalize path length dependent dispersion and attenuation, interpolate any fractional time delays remaining in the backscattered lesion responses after coarse time alignment, and bandpass filter the signal.

The FIR filters can be designed and implemented in either time domain \([8]\) or frequency domain \([10]\). The imaging results presented in this chapter are obtained
using a frequency-domain design procedure [10] and frequency-domain implementation. In this approach, first the time-domain waveforms are transformed to the frequency domain and point-wise-ly multiplied by the frequency-domain beamformer coefficients $W_i[\ell, r_0]$, where $W_i[\ell, r_0]$ is the beamformer weight for the $i^{th}$ antenna location at DFT frequency index $\ell$. The sum of these weighted signals forms the beamformer output $Z[\ell, r_0]$. An inverse DFT transforms the beamformer output back to the time-domain response $z[n, r_0]$. The beamformer coefficients $W_i[\ell, r_0]$ are obtained by solving a penalized least-square problem in the beamformer design stage assuming idealized point scatterers in a homogeneous dielectric medium. Compared to the time-domain FIR filter design approach, frequency-domain filter design is more computational efficient when the number of antenna locations $N$ is large. This is because the frequency-domain approach decouples the filter design across frequencies, circumventing the $NL \times NL$ matrix inversion required in the time-domain design, where $N$ is the number of locations and $L$ is number of FIR taps.

The FIR filter output $z[n]$ is then time-gated again to reduce clutters by ensuring that the output energy is calculated using only samples of $z[n]$ containing backscattered lesion energy. Finally the output energy is calculated for location $r_0$. The reconstructed image of microwave scattering strength is obtained by scanning $r_0$ throughout the reconstruction region and plotting beamformer output energy as a function of location. In this chapter, the 3-D beamformer is designed for the geometry of our experimental setup—a half space filled with soybean oil with $7 \times 7$ array elements with 1-cm lateral spacing placed on top. The 3-D breast region where the beamformer is designed encompasses a $6 \text{-cm} \times 6 \text{-cm} \times 5 \text{-cm}$ domain with a 1-mm pixel resolution.
Figure 33 illustrates the ideal spatial discrimination capability of this 3-D beamformer. The beamformer gain, defined as the output power due to an idealized point scatterer in a homogeneous medium, is plotted on a dB scale as a function of scatterer position of this 3-D beamformer in 3 orthogonal planes cutting through the design location (marked by ‘+’). These patterns indicate that the beamformer attenuates scattered signals originating from any location that is greater than 2 cm away from the design location by more than 10 dB.

5.3 Imaging results

Figure 34 shows the MIST beamforming results for an experimental breast phantom consisting of 4-mm-diameter synthetic tumor placed 2 cm below the skin surface under the center of the array. Tumor simulant #5 is used in this case to illustrate the results of a realistic 5:1 malignant-to-normal tissue contrast. Three orthogonal planes from the 3-D image are labeled using x and y axes that correspond to the lateral dimensions of the imaging domain and a z axis that corresponds to the depth dimension. The origin of the z-axis roughly corresponds to the location of the skin layer. The two energy peaks in the depth direction correspond to scattering from the top and bottom surfaces of the compact cylindrical tumor. The peak energy nearest the surface is located within 2 mm of the top edge of the actual tumor. For comparison purposes, the same beamforming process is also applied to the backscatter waveforms obtained from a tumor-free phantom. The signal-to-clutter ratio (S/C), defined as the ratio of the maximum tumor energy to the maximum clutter energy in the tumor-free phantom, is 14.9 dB.

Figure 35 shows the imaging results for tumor simulant #3 representing a
decreased malignant-to-normal tissue contrast \( \varepsilon_{r, \text{malignant}} / \varepsilon_{r, \text{normal}} = 3.2 \). The S/C is 8.4 dB in this case.

Figure 36 shows the imaging results for a case where the malignant-to-normal tissue contrast is further decreased. Tumor simulant #1 is used in this case to illustrate the results for the most challenging scenario of minimum malignant-to-normal tissue contrast \( \varepsilon_{r, \text{malignant}} / \varepsilon_{r, \text{normal}} = 1.5 \). The S/C is 4.9 dB.

5.4 The influence of dielectric contrast between malignant and normal breast tissue

This section presents a study of the effect of the dielectric contrast between malignant and normal breast tissue on the tumor backscatter response and the image S/C. As explained in Section 5.1, the different dielectric contrasts are created using five tumor simulants with varying dielectric properties.

First, the effect of tissue contrast on the tumor response recorded by a single antenna element is examined using both measurements and simulations. In order to isolate the tumor response, the skin layer is eliminated from the breast phantom. The 4-mm-diameter tumor is placed 3 cm below the antenna in both the numerical and experimental phantoms. The time-domain response is obtained from the simulations and measurements using the same methods described in Section 4.3. The antenna reverberation is removed by subtracting the antenna response obtained with a tumor-free phantom. Figure 37 (a) shows the measured and simulated tumor-response waveform when tumor simulant #5 \( \varepsilon_{r, \text{malignant}} = 5.2 \varepsilon_{r, \text{normal}} \) is used. Note that the peak-to-peak tumor response is about 2 mV when the assumed source waveform has peak-to-peak value of 1.6 V. Thus the time-domain dynamic
range of the system is required to be at least 58 dB to capture this tumor response. Figure 37 (b) plots the peak-to-peak value of the FDTD-computed tumor response as a function of malignant-to-normal tissue contrast. As shown in this plot, the tumor response increases as the tissue contrast increases, as expected. In the case of the minimum dielectric contrast ($\varepsilon_{r,\text{malignant}}/\varepsilon_{r,\text{normal}} = 1.5$), the peak-to-peak tumor response is about 0.4 mV, which requires a 72 dB time-domain dynamic range to be detected.

Additional experiments are conducted with the skin layer present in the breast phantom. Figure 38 (a) shows the measured tumor-response waveforms when five tumor simulants representing the contrast ranging from 5.2:1 (simulant #5) to 1.5:1 (simulant #1). Fig. 38 (b) plots the peak-to-peak values of the tumor responses as a function of contrast in dielectric constants between malignant and normal breast tissue simulants. In the case of the 5.2:1 dielectric contrast, the peak-to-peak tumor response is about 1.3 mV when the peak-to-peak voltage of the source pulse is about 1.6 V. Therefore, a minimum of 62 dB time-domain dynamic range is required to detect this tumor response. In the case of the minimum dielectric contrast ($\varepsilon_{r,\text{malignant}}/\varepsilon_{r,\text{normal}} = 1.5$), the peak-to-peak tumor response is 0.4 mV, which requires a 73 dB dynamic range to be detected. Comparing 37 (b) and 38 (b), we notice slightly smaller tumor responses in the cases when the skin is present. This is because of the loss of transmitted and received microwave energy due to reflection at the skin interface. Note that the dynamic range values quoted here are minimum requirement for a corresponding contrast. In reality, the decrease in malignant-to-normal breast tissue contrast is expected to arise from increase in normal breast tissue density, which is associated with greater attenuation of both transmitted and backscattered microwave signals.
The influence of malignant-to-normal dielectric contrast on image quality and tumor detectability is also studied. MIST beamforming results are obtained for the experimental breast phantom used in Section 5.2 but with five different dielectric properties for the tumor simulants. Image S/C is plotted in Fig. 39 as a function of contrast in $\varepsilon_r$ between normal and malignant tissue simulants. As the contrast increases from 1.5:1 to 5.2:1, the image S/C improves from 4.9 dB to 14.5 dB. The image S/C can be related to tumor response as follows. The tumor response component in the received waveforms contributes to ‘signal’ of the signal-to-clutter ratio, while ‘clutter’ is generated by measurement noise and remnants of the antenna/skin artifacts. Since the clutter components are relatively constant given identical hardware and similar phantom geometry, the increase in tumor response translates directly (but not necessarily linearly) to the improvement in image S/C.
Figure 33: 3-D beamformer gain as a function of position. The orthogonal planes intersect the target position (marked by ‘+’). (a) yz plane at x = 0 cm. (b) xz plane at y = 0 cm. (c) xy plane at z = 3 cm.
Figure 34: Color image of backscattered energy for the multilayer experimental breast phantom, which contains a 4-mm-diameter synthetic tumor located at a depth of 2 cm below the skin surface. The contrast in $\varepsilon_r$ between normal and malignant tissue simulants is 5.2:1. The orthogonal planes intersect the shallower of the two energy peaks of the tumor response. (a) yz plane at $x = 0.1$ cm. (b) xz plane at $y = 0.1$ cm. (c) xy plane at $z = 2.3$ cm.
Figure 35: Color image of backscattered energy for the multilayer experimental breast phantom. The contrast in $\varepsilon_r$ between normal and malignant tissue simulants is 3.2:1. The orthogonal planes intersect the shallower of the two energy peaks of the tumor response. (a) yz plane at x = 0.1 cm. (b) xz plane at y = 0.1 cm. (c) xy plane at z = 2.3 cm.
Figure 36: Color image of backscattered energy for the multilayer experimental breast phantom. The contrast in $\varepsilon_r$ between normal and malignant tissue simulants is 1.5:1. The orthogonal planes intersect the shallower of the two energy peaks of the tumor response. (a) $yz$ plane at $x = 0.1$ cm. (b) $xz$ plane at $y = 0.1$ cm. (c) $xy$ plane at $z = 2.3$ cm.
Figure 37: (a) FDTD-computed and measured tumor response for the case of a 5.2:1 contrast in $\varepsilon_r$ between the malignant and normal breast tissue simulants. The skin layer is eliminated from the breast phantom. (b) Peak-to-peak measure of the tumor response as a function of the contrast in $\varepsilon_r$ between the malignant and normal breast tissue simulants.
Figure 38: (a) Measured tumor responses when the skin is present in the simple breast phantom. (b) Peak-to-peak measure of the tumor response as a function of the contrast in $\varepsilon_r$ between the malignant and normal breast tissue simulants with the skin layer present in the breast phantom.
Figure 39: Image S/C as a function of the contrast in $\varepsilon_r$ between the malignant and normal breast tissue simulants.
Bibliography


Chapter 6

Conclusions

This thesis presents a study of an ultrawideband microwave imaging system for detection of early-stage breast tumors. Our ongoing work in this area is motivated by the clinical need for complementary or alternative modalities to screening X-ray mammography, which suffers from relatively high false-negative and false-positive rates. The physical basis for breast tumor detection with microwave imaging is the contrast in dielectric properties of normal and malignant breast tissues. In the our currently investigated system configuration, an array of antennas is located near the surface of the breast and an ultrawideband (UWB) signal is transmitted sequentially from each antenna. The received backscattered signals are processed using artifact removal and space-time beamforming algorithms to form an image of backscatter energy as function of location. Malignant tumors produce localized large backscatter energy in the image due to their significant dielectric-properties contrast with normal breast tissue. Our detailed numerical and experimental investigation provides the following conclusions:

- In contrast to microwave tomographic approaches that require the solution of a nonlinear inverse-scattering problem, ultrawideband microwave backscatter imaging techniques requires relatively simple, robust space-time beamforming (focusing) techniques. This significant departure from the complicated image reconstruction techniques inherent in conventional tomography is a
consequence of seeking only to identifying the presence and location of strong scatterers, such as malignant tumors, in the breast, rather than attempting to recover the dielectric-properties profile. Our 2-D and 3-D FDTD simulations have demonstrated that a simple delay-and-sum beamforming approach can be used to detect millimeter-sized tumor in fatty breasts.

- We have developed advanced algorithms to adaptively remove the artifact caused by antenna reverberation and skin-breast interface reflection, and improved space-time beamforming techniques to compensate for frequency-dependent propagation effects. Small tumors embedded in heterogeneous normal breast tissue can be successfully detected in a wide range of patient scenarios, even when the contrast between malignant and normal tissue is significantly reduced due to the presence of denser normal breast tissue. Small tumors can also be successfully detected even when a significant mismatch exists between the average normal-breast-tissue dielectric properties assumed in the beamformer design and the actual average dielectric properties of the breast being scanned. Consequently, patient-specific data on the average dielectric properties of normal breast tissue does not appear to be required for detection.

- We report an extensive numerical and experimental investigation of a novel UWB antenna— a pyramidal-horn antenna with a single ridge and a curved launching plane. The antenna has been designed with cm-scale dimensions for low-power UWB microwave radar applications. We demonstrate that the antenna exhibits a very low voltage standing wave ratio over a wide frequency
range from 1 GHz to 11 GHz. This antenna provides high-fidelity transmission and reception of ultrashort microwave pulses with minimal distortion. The spatial distribution of radiated energy has a moderate width of main beam and minimum sidelobes, which is also desirable for biological sensing and imaging applications.

- We present the first experimental demonstration of 3-D MIST beamforming in multilayer breast phantoms with malignant-to-normal dielectric contrasts down to 1.5:1 for a 4-mm synthetic tumor. The enhanced focusing capabilities of MIST beamforming and the efficacy of a data-adaptive algorithm for removing antenna reverberation and reflections from the skin-breast interface are fully demonstrated. The influence of malignant-to-normal breast tissue dielectric contrast on the dynamic range requirements and tumor detectability is summarized using both numerical and experimental results. Our numerical and experimental results suggest that microwave imaging via space-time beamforming offers the potential of detecting small breast tumors using state-of-the-art but readily available hardware and robust signal processing algorithms.