A 3D Active Microwave Imaging System
for Breast Cancer Screening

by

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Dissertation submitted in partial fulfillment of the
requirements for the degree of Doctor of Philosophy
in the Department of Electrical and Computer Engineering
in the Graduate School of
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2008
ABSTRACT

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Abstract

A 3D microwave imaging system suitable for clinical trials has been developed. The anatomy, histology, and pathology of breast cancer were all carefully considered in the development of this system. The central component of this system is a breast imaging chamber with an integrated 3D antenna array containing 36 custom designed bowtie patch antennas that radiate efficiently into human breast tissue. 3D full-wave finite element method models of this imaging chamber, complete with full antenna geometry, have been developed using Ansoft HFSS and verified experimentally. In addition, an electronic switching system using Gallium Arsenide (GaAs) absorptive RF multiplexer chips, a custom hardware control system with a parallel port interface utilizing TTL logic, and a custom software package with graphical user interface using Java and LabVIEW have all been developed. Finally, modeling of the breast (both healthy and malignant) was done using published data of the dielectric properties of human tissue, confirming the feasibility of cancer detection using this system.
# Contents

Abstract iv  
List of Figures viii  
Acknowledgements xii  

## 1 Breast Cancer Screening: Motivation and Goals 1  
### 1.1 Mammography 2  
#### 1.1.1 Breast Imaging Reporting and Data System (BI-RADS) 3  
#### 1.1.2 Harms and Limitations of Mammography 5  
### 1.2 Complementary Methodologies 7  
#### 1.2.1 Ultrasound 7  
#### 1.2.2 MRI 8  
#### 1.2.3 Emerging Methodologies 10  
#### 1.2.4 Biases, Clinical Trials, and Evaluating Efficacy 13  

## 2 Breast Cancer: Anatomy, Histology, and Pathology 16  
### 2.1 Anatomy 17  
#### 2.1.1 Adipose Tissue 17  
#### 2.1.2 Glandular Tissue 18  
#### 2.1.3 Other Tissue 18  
### 2.2 Histology and Pathology 20  
#### 2.2.1 Benign Disorders of the Breast 20  
#### 2.2.2 Cancers of the Breast 23  

## 3 General Theory of Microwave Imaging 29
3.1 Electrical Properties of Human Breast Tissue ....................... 29
3.2 Active Microwave Imaging ........................................... 30

4 Overview of Microwave Imaging Approaches ......................... 32
4.1 Tomographic Approaches ........................................... 33
4.2 Beam Steering Approaches ........................................ 35

5 Beamsteering Imaging Systems ......................................... 39
5.1 MIST Imaging System ............................................... 39
  5.1.1 Skin-Breast Artifact Identification and Removal ............. 39
  5.1.2 Optimal Beamforming ......................................... 41
  5.1.3 Hypothesis Testing ............................................. 46
5.2 TSAR Imaging System ................................................. 48

6 Tomographic Imaging Systems .......................................... 50
6.1 2D Imaging Systems ................................................ 50
  6.1.1 Data Collection Methods and Hardware System Design .... 52
  6.1.2 Forward Modeling and Inversion Methods ................... 57
6.2 3D Imaging Systems ................................................. 62
  6.2.1 Data Collection and System Design ......................... 63
  6.2.2 Forward Modeling .............................................. 64
  6.2.3 Inversion Methods ............................................. 68

7 Antenna Design for Diagnostic Breast Imaging ....................... 70
7.1 General Design Considerations ...................................... 70
7.2 Microstrip Antenna Theory: Analysis and Design ................. 73
  7.2.1 Radiation Mechanism ........................................... 75
<table>
<thead>
<tr>
<th>Section</th>
<th>Title</th>
<th>Page</th>
</tr>
</thead>
<tbody>
<tr>
<td>7.2.2</td>
<td>Theoretical Models of Analysis</td>
<td>77</td>
</tr>
<tr>
<td>8</td>
<td>Microstrip Antennas for Diagnostic Breast Imaging</td>
<td>84</td>
</tr>
<tr>
<td>8.1</td>
<td>Folded Patch Antenna</td>
<td>87</td>
</tr>
<tr>
<td>8.2</td>
<td>Bowtie Patch Antenna</td>
<td>94</td>
</tr>
<tr>
<td>9</td>
<td>System Design</td>
<td>102</td>
</tr>
<tr>
<td>9.1</td>
<td>Switching Network</td>
<td>102</td>
</tr>
<tr>
<td>9.1.1</td>
<td>Multiplexer Design</td>
<td>103</td>
</tr>
<tr>
<td>9.1.2</td>
<td>Switching Network Characterization</td>
<td>105</td>
</tr>
<tr>
<td>9.2</td>
<td>Imaging Chambers</td>
<td>109</td>
</tr>
<tr>
<td>9.2.1</td>
<td>Folded Patch Imaging Chamber</td>
<td>109</td>
</tr>
<tr>
<td>9.2.2</td>
<td>Conformal Arrays and Matching Media</td>
<td>113</td>
</tr>
<tr>
<td>10</td>
<td>Published Imaging Results</td>
<td>127</td>
</tr>
<tr>
<td>10.1</td>
<td>3D Tomographic Images from the Duke MWI Systems</td>
<td>127</td>
</tr>
<tr>
<td>11</td>
<td>Remaining Work and Conclusions</td>
<td>130</td>
</tr>
<tr>
<td>A</td>
<td>Definition of S-Parameters</td>
<td>132</td>
</tr>
<tr>
<td>B</td>
<td>Definition of Antenna Terms</td>
<td>135</td>
</tr>
<tr>
<td></td>
<td>Bibliography</td>
<td>138</td>
</tr>
<tr>
<td></td>
<td>Biography</td>
<td>144</td>
</tr>
</tbody>
</table>
List of Figures

1.1 Sample Mammography Results: (a) Category 1: Negative Finding (b) Category 2: Benign Cyst (c) Category 5: Cancer (Conclusive) [1] . . 3

1.2 Difficulty Imaging Dense Breasts With Mammography. (a) Normal (b)Cancerous [1] . . . . . . . . . . . . . . . . . . . . . . . . . . . . . 7

1.3 Color Enhanced MR Image of the Breast [1] . . . . . . . . . . . . . . 8

1.4 MRI (Right) Clearly Showing Palpable Cancer That Mammogram (Left) Failed to Detect [1] . . . . . . . . . . . . . . . . . . . . . . . . 10

1.5 Thermoacoustic Imaging Process [2]. . . . . . . . . . . . . . . . . . . 12

1.6 Thermoacousting Images of Ductal Carcinoma. Left to right: axial, coronal and saggital views of the cancer (arrows) [2]. . . . . . . 13


2.2 (a) Mild Epithelial Hyperplasia (b) Atypical Epithelial Hyperplasia (c) Carcinoma In Situ (d) Infiltrating Carcinoma (Adapted from [4]) . . 24

2.3 Infiltrating Ductal Carcinoma. (1) Cancer cells surrounding a normal duct. Cancer cells (2) are larger and more irregular than normal cells (3). [5] . . . . . . . . . . . . . . . . . . . . . . . . . . . . . . . . . . . 26


4.1 Simple Delay and Sum Beamformer [6]. . . . . . . . . . . . . . . . . . . 36

5.1 Breast Surface Identification [7]. . . . . . . . . . . . . . . . . . . . . . 40

5.2 Early-Time Artifact Removal Algorithm [7]. . . . . . . . . . . . . . . 40

6.1 Dartmouth Microwave imaging system schematic [8]. . . . . . . . . . 52

6.2 Dartmouth Matching Fluid Formulations [8]. . . . . . . . . . . . . . . 55
<table>
<thead>
<tr>
<th>Section</th>
<th>Description</th>
<th>Page</th>
</tr>
</thead>
<tbody>
<tr>
<td>6.3</td>
<td>Duke 3D Scanning MWI System. [9]</td>
<td>64</td>
</tr>
<tr>
<td>7.1</td>
<td>Patch Antenna with Microstrip Feed.</td>
<td>74</td>
</tr>
<tr>
<td>7.2</td>
<td>Radiation Due to Fringing Fields From “Virtual Apertures” (in Red) on the Ends of the Patch.</td>
<td>75</td>
</tr>
<tr>
<td>7.3</td>
<td>Field Distribution Along Length of Rectangular Patch.</td>
<td>76</td>
</tr>
<tr>
<td>7.4</td>
<td>Far Field Radiation Pattern: (E-plane-Blue, H-Plane-Red).</td>
<td>77</td>
</tr>
<tr>
<td>8.1</td>
<td>Side View of Folded Patch Antenna.</td>
<td>87</td>
</tr>
<tr>
<td>8.2</td>
<td>Top View of Folded Patch Antenna.</td>
<td>88</td>
</tr>
<tr>
<td>8.3</td>
<td>5x5 folded patch antenna array.</td>
<td>89</td>
</tr>
<tr>
<td>8.4</td>
<td>Input Impedance Comparison - Single Antenna vs. Full Panel (HFSS Simulated Results)</td>
<td>90</td>
</tr>
<tr>
<td>8.5</td>
<td>Comparison of Measured and Simulated Return Loss</td>
<td>91</td>
</tr>
<tr>
<td>8.6</td>
<td>Folded Patch Antenna: Field Polarization</td>
<td>91</td>
</tr>
<tr>
<td>8.7</td>
<td>Far Field Radiation Pattern: (E-plane-Red, H-Plane-Black).</td>
<td>92</td>
</tr>
<tr>
<td>8.8</td>
<td>The magnitude of S21 for the transmission (a) between two opposing panels and (b) between front panel and bottom panel.</td>
<td>93</td>
</tr>
<tr>
<td>8.9</td>
<td>Bowtie Patch Geometry: Top View. a = 3mm, b = 9mm, c = 1.25mm d = 2.9mm, e = 9.6mm</td>
<td>95</td>
</tr>
<tr>
<td>8.10</td>
<td>Bowtie Patch Geometry: Side View.</td>
<td>96</td>
</tr>
<tr>
<td>8.11</td>
<td>Imaging applicator using 32 bowtie patch antennas operating at 2.7 GHz.</td>
<td>96</td>
</tr>
<tr>
<td>8.12</td>
<td>Input Impedance Comparison - Single Antenna vs. Full Panel (CST Simulated Results)</td>
<td>97</td>
</tr>
</tbody>
</table>
8.13 Measured return loss for 8 transmitters (Blue), centered about the
CST predicted resonant frequency of 2.7 GHz. (Red) . . . . . . . . . 99
8.14 Bowtie Patch: Local Field Polarization . . . . . . . . . . . . . . . . . 99
8.15 Far Field Radiation Pattern: (E-plane-Red, H-Plane-Black). . . . . . 100
8.16 Sample S21 plot showing a signal about 35 dB between two opposing
panels. . . . . . . . . . . . . . . . . . . . . . . . . . . . . . . . . . . . 101
8.17 Incident signal and scattered signal due to a 1-cm clay ball at the
center of the chamber. . . . . . . . . . . . . . . . . . . . . . . . . . . . . . 101
9.1 Microwave imaging system schematic. . . . . . . . . . . . . . . . . . . 103
9.2 (a) 1x4 Multiplexer (b) 1x16 Multiplexer . . . . . . . . . . . . . . . . 104
9.3 Transmitter multiplexer switching network schematic. . . . . . . . . . 104
9.4 (a) Insertion Loss (b) Antenna Port Reflectivity (c) Channel Isolation 108
9.5 Measured Performance of RF MEMS Switch. Source:www.radant.com 109
9.6 125 element folded patch antenna array. Dark squares indicate transmit-
tting antenna elements and light squares indicate receiving antennas. 110
9.7 Imaging chamber with five-sided cube of antenna arrays surrounded
by microwave absorbing foam. . . . . . . . . . . . . . . . . . . . . . . 111
9.8 Revised Bowtie Patch Imaging Chamber (a) Side View (b) Top View 111
9.9 Clinical Imaging Chamber and Model. . . . . . . . . . . . . . . . . . . 112
9.10 Imaging Chamber Integrated With Alderson Treatment Bra. . . . . . 113
9.11 Properties of Acetone (a) Measurement Apparatus (b) $\varepsilon_r$ (c) $\sigma$ . . . . . . 114
9.12 Microwave Imaging Cart for Clinical Use. Front view shows the central
applicator for the imaged breast, and space for either contralateral
breast. Side view shows the imaging chamber, head and arm rest, leg
gap, and cabinet for control and measurement electronics. . . . . . . 116
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Chapter 1

Breast Cancer Screening: Motivation and Goals

In the fight against breast cancer, early detection through screening is the best tool short of a complete cure. Breast cancer screening is defined as “the evaluation of a population of asymptomatic women, who have no overt signs or symptoms of breast cancer, in an effort to detect unsuspected disease earlier in its growth” [10]. By detecting cancer in its early stages, it can be identified and treated before it has the opportunity to spread and become potentially lethal.

According to a report by the U.S. Institute of Medicine (IOM) in 2001, the ideal screening system would have the following properties: it should be noninvasive, provide minimal discomfort and minimal health risk; furthermore, it should be able to specifically detect malignant tumors at the earliest possible stage, all while being cost effective, easy to perform, and provide conclusive, consistent results.

While x-ray mammography currently offers the best combination available of these ideal characteristics, other established imaging techniques, such as ultrasound and magnetic resonance imaging, have been used with some success to supplement mam-
mography in certain specific cases. For example, ultrasound can be used to determine whether a lesion detected by a mammogram is a liquid cyst or a solid tumor. However, even with the combined use of mammography, ultrasound, and MRI techniques, the current method of screening for breast cancer does not meet the ideal requirements of the IOM report. Therefore, researchers are actively searching for alternative modalities of screening and diagnostic breast imaging.

1.1 Mammography

In the first fifty years (1940-1990) since statistics began to be kept on breast cancer incidence and death rates in the United States, breast cancer mortality rates remained roughly the same. Interestingly, in 1990, the death rate suddenly began a steady decline. By 2005, the mortality rate had decreased by a full 25 percent. To understand why this happened, another major statistical change must be examined: in 1983, the widespread adoption of screening mammography resulted in a sharp increase in the number of breast carcinomas diagnosed in the United States. However, as periodic screening with mammograms is not effective in preventing deaths from aggressive cancers, there was not an immediate effect on mortality rates. Still, early detection and treatment of moderately growing cancers began to influence the mor-
tality rate statistics starting in 1990. The ensuing steady and sustained decline in breast cancer deaths, along with the conclusive results of several randomized clinical trials showing statistically significant reductions in mortality, has proved that breast cancer screening saves lives [10].

1.1.1 Breast Imaging Reporting and Data System (BI-RADS)

![Sample Mammography Results: (a) Category 1: Negative Finding (b) Category 2: Benign Cyst (c) Category 5: Cancer (Conclusive)](image)

**Figure 1.1** Sample Mammography Results: (a) Category 1: Negative Finding (b) Category 2: Benign Cyst (c) Category 5: Cancer (Conclusive) [1]

The Breast Imaging Reporting and Data System (BI-RADS), developed by the American College of Radiology, established a common lexicon of terminology, standardized assessment categories, and a simple coding system for breast imaging results. The system was designed to improve clarity of communication and decision-making with regard to breast imaging, as well as to facilitate the analysis of the outcome data.
to improve the quality of screening and patient care. Despite being initially developed for mammography, it has since been also adapted for MRI and ultrasound results, and provides an excellent framework for evaluating the results of any breast imaging methodology. A brief description of the different assessment categories follows [11]:

0: Incomplete. This finding is used whenever the imaging result is inconclusive and further evaluation (previous or additional mammograms, ultrasound, etc.) is needed.

1: Negative. The vast majority of screening mammograms are in this category, and no further action is required. The breasts are symmetrical and no masses, architectural disturbances or suspicious calcifications are present.

2: Benign finding(s). This is also a negative mammogram, with a definitive conclusion from imaging that a benign condition exists. For example, certain fibroadenomas, cysts, lipomas, and hamartomas all have characteristic appearances, and may be labeled with confidence. Other findings, such as intramammary lymph nodes or implants may be noted, while concluding that there is no evidence of malignancy.
3: Probably benign. Findings in this category have a very high probability of being benign. It is not expected to change over the follow-up interval, but the radiologist would prefer to establish its stability.

4: Suspicious abnormality. This category includes all lesions that do not show the classic visual signs of breast cancer but have a high enough probability of being malignant that a biopsy is recommended.

5: Highly suggestive of malignancy. The imaging results clearly show signs of malignancy, and one-stage surgical treatment should be considered without preliminary biopsy.

6: Known biopsy:proven malignancy. This category is for imaging studies that take place after malignancy has already been confirmed.

1.1.2 Harms and Limitations of Mammography

Despite the fact that mammography has proven effective in saving lives, there is potential for harm associated with any screening methodology. Beyond the direct monetary and time costs associated with testing a large population of women, there are a number of other potential negative effects that arise from screening. Because
screening is performed, by definition, on apparently healthy women, it is often the source of unnecessary anxiety and pain, especially due to inconclusive or slightly abnormal results. Frequently, biopsies will be performed only to return a benign finding. The greatest harm, however, results from the detection of cancers that would not have affected the patient had they remained undiscovered. Finally, and most tragically, is the harm associated with the discovery and subsequent treatment (surgery, radiation, chemotherapy, etc.) of cancers that prove to be just as lethal as they would have been if left untreated [10].

Beyond the above potential harms, the IOM’s 2001 report highlighted a number of serious shortcomings in conventional mammography: uncomfortable breast compression discourages some patients, difficulty imaging dense breasts reduces imaging sensitivity (see Figure 1.2), false positives lead to unnecessary biopsies (roughly 10 percent of mammograms contained images with possible tumors, but less than one in ten of those were diagnosed as malignancies), and most importantly, mammography misses up to 15 percent of actual tumors, some of which were even detectable by palpation [12]. All of these limitations and their associated potential for harm, along with the additional health risks associated with x-ray’s ionizing radiation, provide
ample motivation for the development of alternative forms of breast imaging.

![Figure 1.2 Difficulty Imaging Dense Breasts With Mammography. (a) Normal (b)Cancerous [1]](image)

### 1.2 Complementary Methodologies

#### 1.2.1 Ultrasound

Ultrasound imaging (also called sonography) relies on high frequency sound waves that reflect with varying intensity from different tissues. In the breast, ultrasound is able to differentiate skin, fat, glandular tissue, and muscle. However, for a number of reasons, ultrasound has yet to prove itself as an effective breast cancer screening methodology, despite improvement in recent years. A major limitation of ultrasound, unfortunately, is that breast fat and most cancer cells have similar acoustic properties, which makes detecting many tumors impossible with ultrasound. Also, it is important to recognize that the vast majority of ultrasound procedures are performed using
hand-held devices, making the imaging results highly operator-dependent. As a result, ultrasound primarily remains a tool for distinguishing cysts from solid tumors and for guiding biopsy procedures. Until new techniques are developed that resolve these and other issues, ultrasound cannot be considered as an effective widespread screening method for breast cancer.

1.2.2 MRI

![Color Enhanced MR Image of the Breast](image)

**Figure 1.3** Color Enhanced MR Image of the Breast [1]

Magnetic resonance imaging relies on the interaction of RF energy and strong magnetic fields with the magnetic properties of certain atoms to produce high resolution images. In MR imaging, a strong magnet is first used to align the protons of the
hydrogen (also possibly phosphorus or sodium) atoms, and then pulsed RF energy is used to tip the protons out of alignment. Once the RF pulse ends, as the individual protons return to alignment (spin lattice or T1 relaxation) and begin to precess at different rates (spin-spin or T2 relaxation), they emit an RF signal that is detected by antennas in the MR device. Imaging is possible because these return signals vary in phase and intensity based on the strength of the magnetic field, the frequency and pattern of the RF pulses, and the properties of the tissue. For example, breast cysts and other tissue that contains “free water” have long T2 relaxation times, whereas glandular tissues containing “bound water,” with its associated greater intermolecular interaction, tend to have shorter T2 relaxation times. By encoding location information using magnetic field gradients and different sequences of RF pulses, the tissue properties at each location can be mapped to form high resolution 3D images.

A variety of studies [13, 14] have shown that MRI with IV injected contrast agent successfully detected many breast cancers that were otherwise undetectable by mammogram (see Figure 1.4), ultrasound, or clinical breast exam. However, MRI also suffered from a higher false positive rate than any of the other screening methods. In one study [10], MRI had a specificity of only 95.4 percent, compared to 96, 99.3, and
99.8 percent for ultrasound, clinical breast exam, and mammography, respectively.

This low specificity, along with the high costs associated with MR imaging and the need for IV contrast, severely limit MRI’s usefulness as an alternative or complement to mammography as a widespread screening method.

![Figure 1.4 MRI (Right) Clearly Showing Palpable Cancer That Mammogram (Left) Failed to Detect][1]

### 1.2.3 Emerging Methodologies

**Optical Coherence, Electrical Impedance, and Microwave Imaging**

**Optical Coherence Tomography (OCT).** Optical coherence tomography uses interferometry of near-infrared light waves to create high-resolution images. OCT works by taking advantage of the fact that only non-scattered light will maintain co-
herence and create interference patterns when added to a reference signal, provided the two signals have traveled the same distance (within the coherence length). Scanning is then made possible by varying the optical length of the reference arm. While OCT does not offer enough depth of penetration to perform full scans of the entire breast, it has the potential to improve biopsy results by providing real-time optical images using fiber optic catheters.

**Electrical Impedance Tomography (EIT).** In EIT, multiple electrodes are either applied directly to the skin or coupled to the skin using a conductive medium. A small AC electric current, at a frequency in the kHz range, is applied across two or more electrodes, while the remaining electrodes are used to measure voltage. This procedure is then repeated for numerous combinations and/or patterns of applied current. An inversion algorithm is then applied to the measured data in order to solve the ill-posed, non-linear inverse problem and create an image of the breast.

**Microwave Imaging (MWI)** In a process similar to EIT, MWI uses antennas to measure the interaction of electromagnetic waves (generally between 1 and 5 GHz.) with tissue of varying electrical properties. Image reconstruction can be performed
using tomographic methods similar to those described above for EIT, or by using beam forming techniques (similar to RADAR) in order to scan the volume of the breast point-by-point. Both of these methods will be described in much greater detail in later chapters.

Thermoacoustic Imaging

Figure 1.5 Thermoacoustic Imaging Process [2].

Thermoacoustic imaging is a very promising hybrid technique that combines elements of microwave and ultrasound imaging methods. As shown in Figure 1.5, thermoacoustic waves are generated in a multistage process. First, the tissue is irradiated by pulsed electromagnetic radiation, typically microwaves in the case of breast
imaging. The energy from the microwaves is absorbed and converted to heat, which raises the temperature of the tissue very slightly, causing the tissue to expand in volume. This expansion produces an acoustic wave that propagates outward in all directions from the site of energy absorption. These acoustic waves (which are typically ultrasonic) are then detected by an array of transducers and used to reconstruct images such as shown in Figure 1.6.

![Figure 1.6 Thermoacousting Images of Ductal Carcinoma. Left to right: axial, coronal and sagittal views of the cancer (arrows) [2].](image)

### 1.2.4 Biases, Clinical Trials, and Evaluating Efficacy

When evaluating any of these emerging screening methods, it is important to keep in mind that proving any screening method’s efficacy is not a straightforward task. The process of evaluating breast cancer screening systems in particular is complicated by the fact that breast cancer is not usually a quickly lethal form of cancer. This leads to the potential for the introduction of a number of biases in any screening study, and makes it difficult to demonstrate conclusively that the screening method has led
to a reduction in mortality.

**Lead-time Bias.** Lead-time bias occurs when earlier detection is falsely attributed to longer survival time. This is due to the fact that survival time is calculated from the time of diagnosis, rather than the unknowable time when the cancer first formed. As a result, earlier diagnosis automatically results in statistically longer survival times, even if the life of the patient has not actually been prolonged.

**Length-bias.** This is a sampling bias that results from the fact that periodic screening is more likely to detect slower-growing (and less lethal) forms of cancer. This apparent reduction in the death rate from cancers that were detected by screening could be falsely attributed to earlier detection of cancer in general.

**Selection Bias.** Selection bias results from the fact that women who are willing to participate in a screening trial are more likely to be in good health than those who do not self-select for screening.

**Pseudodisease Bias.** This bias results from the fact that screening detects many cancers that would be nonlethal even if they remained undetected (pseudodiseases).
Patient survival that is incorrectly attributed to detection of these cancers therefore inflates the effect of screening on mortality rate.

**Cause-of-Death Bias.** This bias results from the natural inclination of those holding hammers to confuse all of their problems for nails. The cause-of-death in patients in screening trials with multiple diseases may be unknowingly biased to fit the desired outcome of the trial [10].

**Randomized Controlled Trials.** Given all of the biases listed above, properly designed randomized controlled trials (RCTs) are the only conclusive method of proving that a screening methodology results in a reduced mortality rate. In RCTs, a statistically large group is randomly divided into two subgroups, with one being offered screening while the other acts as a control. The goal is to use the laws of probability to eliminate the effect of biases on the results of the study. Only when a screening methodology has successfully demonstrated a statistically significant reduction in mortality through a randomized controlled trial can it be stated with any certainty that an effective alternative to mammography has been found.
Chapter 2

Breast Cancer: Anatomy, Histology, and Pathology

In the development of an alternative modality of breast cancer screening, it is important to have a basic understanding of the anatomy, histology, and pathology of the breast.

2.1  Anatomy

The breast (see Figure 2.1) is a modified skin gland that lies on the chest wall, usually between the clavicle and the sixth rib, and is bounded externally by skin and internally by the pectoralis muscle. Breast tissue also extends up into the axilla (underarm region) via a pyramidal-shaped axillary tail. The tissue in the breast primarily consists of a combination of fat and glandular tissue, with the relative proportions of the two varying widely. The remainder of the breast is made up of connective tissue, vascular tissue, lymphatics, and nerves. The key features of each of these elements are described below.

2.1.1  Adipose Tissue

The adipose tissue in the breast can be divided into three main groups: subcutaneous (directly beneath the skin), retromammary (back of the breast), and intraglandular (between the glandular structures). The subcutaneous and retromammary fat regions form a layer between the majority of the glandular tissue and the external boundaries (skin and pectoralis muscle) of the breast. The majority of cancers develop in the region of glandular tissue within 1 cm of these fat layers. Unfortunately, the layers of fat do not completely isolate the glandular tissue from either the surrounding skin
or muscle, which enables the cancer to spread. The amount of fat generally increases with age, body mass, and breast size, but this is by no means an absolute [10].

2.1.2 Glandular Tissue

The glandular tissue in the breast consists of a number of discrete lobes, which are made up of lobules (which produce milk) and ducts (which deliver it to the nipple). There are approximately 15-20 lobes in each breast, and each lobe is thought to be exclusively drained by its own individual duct system. Within a lobe are dozens of lobules 2-3 mm in diameter, and within each lobule are as many as 100 alveoli (often referred to as acini), which are the basic secretory units of the breast. The ductal system can be thought of as a tree-like structure, with a single lactiferous duct that opens at the nipple acting as a trunk that branches into many smaller ducts that each end in what has been termed the terminal duct lobular unit (TDLU) [15].

2.1.3 Other Tissue

Skin and Connective Tissue. The breast is supported by a combination of connective tissue and skin. The skin overlying the breast generally varies in thickness between 0.8mm and 3mm, and skin thickness tends to be inversely proportional to breast size. The skin also contains the nipple, which is slightly below the centerpoint
of the breast and extends about 5-10mm above the skin surface, and the surrounding areola, which contains a number of small bumps called Montgomery’s glands, which lubricate the skin during breast feeding. Finally, the breast is supported by a surrounding layer of fascia (connective tissue), which is interspersed with the subcutaneous and retromammary adipose tissue, and by a number of suspensory ligaments (Cooper’s ligaments) that provide an internal supporting framework for the breast lobes [15].

**Vascular and Nerve Tissue.** The breast receives its blood supply from a variety of axillary, internal thoracic, and intercostal arteries. Each artery generally has a corresponding veinous channel which drains a web of veins originating from the nipple, the subcutaneous layer, and the glandular tissue of the breast. Similarly, the breast receives its nerve supply from multiple branches of the intercostal nerves [16].

**Lymphatic Tissue.** The lymphatics, vein-like vessels that drain lymph fluid into the blood stream, represent the body’s main line of defense in the body’s immune response, but they also provide a conduit for cancer cells to spread and metastasize. In the breast, the vast majority of lymphatics travel from the nipple to lymph nodes
in the axilla, and from there to nodes above the collar bone. Some lymphatics also

drain the rear part of the breast to nodes under the breast bone (internal thoracic

nodes). Finally, there are also a small number of lymph nodes within the breast itself

[17].

2.2 Histology and Pathology

Under normal, healthy, conditions, both the ductal and lobular cells in the breast are

lined with a specialized two-layer cell lining that is attached to a basement membrane.

The inner layer is made up of epithelial cells that perform secretory and absorptive

functions, while the outer layer is made up of contractile myoepithelial cells [16]. It

is in this cell lining where cancer is thought to originate, and the status of these cells

is often critical in distinguishing benign diseases from malignancy.

2.2.1 Benign Disorders of the Breast

Approximately 90 percent of lumps and suspicious breast lesions turn out to be one

of a variety of benign breast disorders. These disorders include breast cysts, adeno-

mas, fibroadenomas, papillomas, fat necrosis, mammary duct ectasia, and epithelial

hyperplasia. A brief description of some of the more common benign conditions

follows.
Cysts, Fibroadenomas, and Adenomas. Cysts are found throughout the body and are generally fluid-filled sacs that are lined with cells that actively produce secretions that fill the cyst. Breast cysts are different in that they are in fact closed-off portions of mammary ducts that passively fill with fluid. Breast cysts are round, smoothly contoured, and are generally movable [17]. Often, they are detected by mammogram or palpation, but are later diagnosed by ultrasound or biopsy.

Like cysts, fibroadenomas present as round, smoothly contoured, movable lumps. These benign tumors are made of epithelial and stromal cells, which gives them a firm, but not hard, consistency. They are generally solitary (in 80 percent of cases), and rarely exceed 3cm in diameter. Fibroadenomas have been shown to have malignant changes in about 0.1 percent of cases (mostly in the form of LCIS) [16].

True adenomas (or tubular adenomas) are neoplastic tumors that are composed of tubular structures lined by a single layer of epithelial cells. They tend to form sharply circumscribed nodules roughly 2 cm in size. Adenomas occur much less frequently than fibroadenomas, but are more likely to be mistaken for cancer in screening. Biopsy showing the presence of actin is used to distinguish this benign tumor from cancer [16].
Papillomas and Mammary Duct Ectasia. Papillomas and mammary duct ectasia are disorders of the ducts that often lead to nipple discharge. Intraductal papillomas are small polyp-like growths in the ducts directly behind the nipple which often cause a watery, bloody discharge from the nipple. Mammary duct ectasia is a disorder that causes the ducts to become distended and clogged, often causing a lump, swelling, and nipple discharge. While these conditions cause discharge, one of the signs of malignancy, both papillomas and mammary duct ectasia are benign conditions.

Fat Necrosis. Fat necrosis is a condition, commonly associated with trauma, that occurs when a group of cells in the breast dies, leaving behind a small, hard, flat lesion. Although necrosis is often a sign of malignancy, this condition is distinguished from carcinoma by its yellow color and the bulge it typically forms in the adipose tissue [16].

Epithelial Hyperplasia. Epithelial hyperplasia is the generic term for the proliferation of the epithelial cells that line the ducts and lobules in the breast. Epithelial hyperplasia can be classified by its severity: mild (three to four cells thick), mod-
erate (more than four cells), florid (completely filling the lumen), and atypical (any proliferation that shows signs of potential malignancy). Hyperplasia in the breast is distinguished from neoplasia (the process underlying malignancy) by the following features: uniformity of the nuclei, the presence of myoepithelial cells, and absence of necrosis. When additional conditions are present alongside epithelial hyperplasia (such as papilloma, mammary duct ectasia, or fat necrosis), it can become diagnostically difficult to distinguish from carcinoma, but the presence of myoepithelial cells and uniform nuclei would still suggest benignancy [16].

Although epithelial hyperplasia is a benign condition, studies have shown an association with an increased risk for the development of carcinoma. Patients with moderate and florid hyperplasia have a slightly increased risk (1.5-2 times), and those with atypical hyperplasia are five times more likely to develop carcinoma [16].

2.2.2 Cancers of the Breast

Breast cancer comes in many forms, and is distinguished by the type of tissue where the malignancy originates and whether the cancer is confined to that original site. If the cancer originates in the lobules that manufacture milk or the ducts that carry it, as most commonly occurs, it is classified as carcinoma of the breast. Cancers
that originate in the muscle, fat, or connective tissue in the breast are referred to as sarcomas. If the cancer is confined to its original site, and has not yet formed a tumor mass, it is referred to as \textit{in situ} cancer, otherwise, it is considered to be infiltrating. Furthermore, if the cancer has spread beyond the breast and its associated lymph nodes to distant parts of the body, it is said to have metastasized.

There is considerable debate over whether breast cancer develops in a continuous progression from epithelial hyperplasia to carcinoma \textit{in situ} and finally to infiltrating carcinoma (see Figure 2.2), or if each condition develops independently. Although there is considerable evidence to support the theory of progression, to date there has been no scientific proof. In any case, studies have shown conclusively that those with atypical hyperplasia and DCIS are at increased risk for invasive carcinoma.

\begin{figure}[h]
\centering
\includegraphics[width=\textwidth]{figure2.png}
\caption{(a) Mild Epithelial Hyperplasia (b) Atypical Epithelial Hyperplasia (c) Carcinoma \textit{In Situ} (d) Infiltrating Carcinoma (Adapted from [4])}
\end{figure}

\textbf{Ductal Carcinoma \textit{in situ}.} Ductal Carcinoma \textit{in situ} (DCIS) comes in many forms, which are classified based upon the patterns the cancer cells form. As the
cancer cells lining the ductal wall proliferate, they begin to fill the inner space of the duct: first with small protrusions (micropapillary DCIS), then larger ones (papillary DCIS); sometimes the protrusions form a lattice structure (cribriform DCIS) and in others the duct is entirely filled (solid DCIS). In the most severe cases, the center cells die, forming a necrotic core (comedo DCIS) [17].

**Lobular Carcinoma in situ.** In Lobular Carcinoma in situ (LCIS), the lobules are completely filled by round, uniform, medium-sized cells. Occurrences of atypia, pleomorphism, and necrosis are also found infrequently in LCIS. Since LCIS cannot be detected by palpation and is unlikely to appear on a mammogram, it is usually discovered as an incidental finding in breast tissue that has already been removed due to suspicion of other disease. Statistically, LCIS is bilateral in about 30 percent of cases and the majority of cases are found within 5 cm of the skin surface near the nipple [16].

**Infiltrating Carcinomas**

Of the carcinomas of the breast, the vast majority (85 percent) form in the ducts, while the remaining fifteen percent form in the lobules. Since the majority of ductal and lobular cells are found in the upper part of the breast, the center of the
breast, or in the outer quadrant, this is also where most cancers occur. Other, less common, invasive carcinomas include tubular carcinoma, mucinous carcinoma, and medullary carcinoma, all of which have better prognoses than the ductal or lobular types. Tubular carcinoma is usually small in size (about 1 cm in diameter) and is similar in appearance to adenomas but lacks myoepithelial cells, which is confirmed by a negative finding for actin after biopsy. In mucinous carcinoma, the tumor appears as a well-demarcated mass composed of tumor cells floating in pools of mucin. Medullary carcinoma is characterized by a solid mass with a well-defined border that contains diffuse clusters of large, pleomorphic tumor cells that have been infiltrated
by lymphocytes. Finally, invasive carcinomas may be termed inflammatory, which describes the condition that occurs when the cancer has extensively spread to the dermal lymphatics, causing edema, tenderness, and swelling of the breast. The prognosis of inflammatory carcinomas is generally quite poor [16].

**Infiltrating Ductal Carcinoma.** Infiltrating ductal carcinoma is the most common type of breast cancer, accounting for about 75 percent of all breast cancer [18], and is the most common lethal cancer of the breast [10]. In this invasive form of breast cancer, the malignant cells break through the basement membrane of the mammary duct into the surrounding fatty tissue. This leads to the growth of fibrous, scarlike tissue around the cancer cells, which can be detected both on a mammogram and by palpation [17].

**Infiltrating Lobular Carcinoma.** Unlike the ductal form, this invading carcinoma does not provoke the growth of fibrous tissue and tends to infiltrate in a more diffuse manner, forming single file groupings of malignancy in the surrounding tissue. As a result, it is much more difficult to detect on a mammogram and often feels more like a thickening of tissue rather than a lump. Unfortunately, this means that it is
usually detected at a later stage than its ductal counterpart.

**Sarcomas and Other Rare Types**

There are a variety of sarcomas (all rare) that may occur in the breast, including angiosarcoma (sarcoma of blood or lymphatic vessels), fibrosarcoma (sarcoma of connective tissue), liposarcoma (sarcoma of adipose tissue), osteosarcoma (bone forming sarcoma), and chondrosarcoma (cartilage forming sarcoma). One of the more common sarcomas found in the breast is Phyllodes tumor (formerly called cytosarcoma Phyllodes), which mostly tends to affect younger women. Phyllodes tumors can grow as large as 15cm and are similar in appearance to fibroadenomas. These tumors often recur locally, but rarely metastasize. Finally, breast tumors can also form as a result of other cancers such as lymphoma and leukemia, or due to metastases from cancers such as melanoma and lung cancer; however, all of these are very rare [16].
Chapter 3

General Theory of Microwave Imaging

The fundamental basis of any breast imaging system is the contrast that exists between the material properties of healthy and malignant tissue. As mentioned previously, the current methods commonly used in clinical applications rely on contrast in density (x-ray), acoustic impedance (ultrasound), and nuclear spin relaxation (MRI). However, an additional source of measurable contrast arises from the different electrical properties of healthy and malignant breast tissue.

3.1 Electrical Properties of Human Breast Tissue

The theoretical basis for microwave breast imaging is the high contrast between the permittivity and conductivity of healthy and malignant breast tissue. This contrast has been documented in a variety of studies [19, 20, 21, 22, 23, 24].

Figure 3.1 shows graphs of conductivity and relative permittivity of normal and malignant breast tissue over the microwave frequency range. While there is variation among the values reported by the different research groups, all studies show substantial contrast. This suggests that microwave imaging is theoretically possible.
at many frequencies, and that the choice of frequency is largely a matter of balancing the added spatial resolution afforded by higher frequencies with the additional attenuation that comes with it. This has led to the investigation of a variety of imaging systems based on electrical property contrasts. These systems include the aforementioned microwave-induced thermoacoustic imaging, EIT, and MWI systems.

3.2 Active Microwave Imaging

In general, active microwave imaging falls under the broad category of inverse problems. The objective of MWI is to reconstruct the unknown distribution of the complex permittivity inside the mammary tissue given a set of data measured along the perimeter of the volume. There are several groups currently performing research in active microwave imaging [6, 25, 9, 26, 27, 28, 29, 30].
While these systems take varying approaches to the specifics of data acquisition and image reconstruction, they share the common potential advantages that microwave imaging offers over traditional mammography. First, active microwave imaging systems do not require the use of ionizing radiation. In addition, MWI systems would eliminate the need for uncomfortable breast compression. Furthermore, microwave imaging systems should be less expensive than x-ray systems and much less costly than MRI. All of these characteristics would allow for earlier and more frequent examinations. Finally, microwave imaging has the potential for a much higher sensitivity in detecting tumors, as well as higher specificity in differentiating malignant and benign tumors. This is due to the fact that the electrical contrast is likely to be much higher than the density contrast, particularly for malignant lesions.
Chapter 4

Overview of Microwave Imaging Approaches

Although research groups have taken a variety of approaches to microwave imaging, these approaches can be divided into two main categories: tomographic methods that utilize traditional inverse scattering approaches (both sequential 2D slice methods and full 3D methods) and methods that use beamsteering approaches (confocal imaging, MIST, TSAR). The fundamental difference between these two methods is the way in which a voxel of material data is localized. In the case of inverse scattering approaches, the electrical properties of all locations in the volume are determined simultaneously using an inversion algorithm that operates on the aggregate of the measured data. Alternatively, beamsteering approaches isolate a particular location in the volume to be measured, and data is taken for each individual voxel. Statistical methods are then employed to determine a given voxel’s electrical properties. Further details of each of these methods are described below.
4.1 Tomographic Approaches

In general, tomographic microwave imaging systems consist of array of narrowband antennas that are either mechanically scanned or electronically switched to record a set of scattering parameters on or near the surface of the breast. This entire set of data is then operated on in an iterative fashion using an inversion algorithm that seeks to reconstruct a complete map of the actual dielectric properties of the breast. In order to accomplish this, an ill-posed, non-linear inverse scattering problem must be solved.

Sequential 2D Slice Approach

In the case of the 2D slice approach, the volume to be scanned is surrounded by 2D array of antennas, typically arranged in a ring, and sequential 2D images are generated by translating the entire array vertically through a series of positions. There are several advantages to using such an approach. First, a relatively small number of simple antennas can be used to illuminate the breast, significantly reducing the complexity of the system to be modeled in the reconstruction algorithm. Furthermore, the 2D nature of the problem itself further simplifies the inversion process. In addition, a translating 2D array reduces the complexity of the hardware system design and data
acquisition process. Difficulties associated with packing a dense number of antennas in a full 3D configuration are eliminated, and isolating sensitive electronics from the necessary matching fluid is also much easier to accomplish. Finally, the positioning of the antennas is potentially more flexible since their locations are determined by the CNC motor.

This approach is not without its disadvantages, however. Due to the nature of the motorized translation system, data acquisition time will be much slower than an electronically switched 3D imaging system. This prolonged acquisition time, combined with the motion associated with the translating antenna array (disturbing both the matching fluid level and possibly altering the position of the breast itself), will likely lead to increased noise. Also, a 2D approach to data collection reduces the number of possible antenna combinations and does not have the ability to utilize data from out of plane antenna combinations. Finally, such a 2D approach is likely to result in reduced resolution and errors since the translation in the transverse direction is on the order of a wavelength. For a further discussion of the tradeoffs associated with 2D imaging, the reader is referred to [31].
Full 3D Inversion Approach

The full 3D inverse scattering approach offers many potential advantages over the other two approaches detailed in this paper, often at the cost of added complexity. First, a full 3D array of densely packed antennas that are switched electronically offers efficient data collection and highly precise antenna positioning. A complete 3D array also provides the maximum combination of transmit and receive antennas, and enables the collection of out-of-plane transmission data. Finally, a fully 3D system is able to take advantage of the optimizations and advances in 3D inversion techniques that have recently been made. All of these contribute to the potential of 3D inversion methods to offer greater resolution, increased SNR, and more rapid data collection [32, 33, 34].

4.2 Beam Steering Approaches

Beamforming systems have their roots in optics, confocal microscopy, and RADAR. The general approach in such systems is to focus an illuminating microwave signal at a particular point in the scanning volume and then to refocus the scattered signal back to the point of illumination. By systematically scanning the focal point within a set of preselected voxels throughout the breast volume, a 3D image can be constructed.
The ability of these systems to detect tumors relies on the increased backscatter caused by malignant tissue. The primary advantages of this approach are two-fold: there is no need for complex inversion techniques and simple time-gating methods can be used to reduce clutter and multiple scattering effects.

Confocal Imaging: Simple Delay-and-Sum Beamforming

Figure 4.1 Simple Delay and Sum Beamformer [6].

In a typical confocal breast imaging system, ultrawideband (UWB) pulses are generated by an antenna located on or near the surface of the breast. The backscattered waveform at that particular antenna location is then collected and stored in a computer. Using both electronic switching and mechanical scanning, this procedure is repeated for each antenna element in an N element antenna array. The set of N backscattered signals are then time-shifted to achieve coherent addition for a specific
focal point within the breast. The focal point is then scanned throughout the breast by adjusting the relative amount of time-shift applied to each backscattered signal. Statistical analysis is then used to predict the absence or presence of a tumor at each scanned voxel. Finally, all of the statistical decisions for each voxel are aggregated to form an image. It should be noted that this method does not seek to generate a map of the dielectric properties of the breast. Instead, it only seeks to identify and locate the presence of strong scatterers within the breast [35].

**Microwave Imaging via Space-Time Beamforming**

Since the simple delay-and-sum beamformer described above does not have the ability to compensate for dispersion and other frequency-dependent propagation effects, Davis et al. developed a microwave imaging via space-time (MIST) beamforming system. Like the confocal system, each element of an antenna array both transmits an UWB pulse into the breast of a patient lying in the supine position and measures the backscattered waves. Just as in the delay-and-sum beamformer, the MIST beamformer performs spatial focusing by time-aligning these backscattered waves. Additionally, however, the MIST beamformer compensates for frequency-dependent propagation effects and optimally discriminates against artifacts and noise. This is
achieved through the use of a frequency dependent filter in each channel, the details of which will be described in a later section [7].

**Tissue Sensing Adaptive Radar**

As an alternative to MIST, Fear and Sill have developed a similar system that has been termed Tissue Sensing Adaptive Radar. Like MIST, it seeks to address the shortcomings of the simple confocal system; however, there are a few fundamental differences between the two. For example, in the TSAR system, the patient lies prone and the pendulous breast is scanned from surrounding locations. Additionally, the TSAR system uses less complicated clutter reduction methods than MIST. Further details of the TSAR system and its differences with the MIST system will be described in a later section.
Chapter 5

Beamsteering Imaging Systems

5.1 MIST Imaging System

The MIST imaging system can be divided into three main processing stages: pre-processing that eliminates the large reflection at the skin, optimal beamforming, and hypothesis testing. Details of each of these stages are described below.

5.1.1 Skin-Breast Artifact Identification and Removal

Because the skin layer has vastly different dielectric properties from the underlying mammary tissue, the reflected signal from the surface of the breast is significantly larger than signal from any potential tumors inside the breast. Therefore, it is essential that this signal be identified and removed.

Breast Surface Identification Procedure

1. Apply matched filter to backscattered signal in each channel to estimate the propagation time to the skin. This localizes the skin to a spherical surface surrounding the antenna.

2. Assume the skin surface is convex and tangent to the surface of each sphere.
3. Estimate the surface by assuming that neighboring spheres share the same tangent.

Figure 5.1 Breast Surface Identification [7].

Figure 5.2 Early-Time Artifact Removal Algorithm [7].

Early-Time Artifact Removal (Skin Subtraction)

The removal of the large backscattered signal (early-time artifact) received from the skin-breast interface is accomplished by subtracting an estimate of the artifact made up of a filtered combination of the signals received at all other antenna locations as
shown in Figure 5.2. The filter weights used to generate this estimate are chosen to minimize the residual mean-squared error over the early-time response (where the artifact dominates). This procedure is repeated for each antenna in the array. Notably, this method does not require any *a priori* knowledge of the skin interface or the underlying breast tissue as it depends solely on the measured data.

### 5.1.2 Optimal Beamforming

Following the skin-breast artifact removal, the signals are then passed through the MIST beamformer. The six main processing steps of the MIST beamformer are as follows:

1. Time alignment for a specific scan location, \( r_0 \).
2. Time-gating to reduce interference and clutter.
3. Transform to the frequency domain and multiply by the beamformer coefficients.
4. Sum weighted signals.
5. IDFT to return signals to the time domain where time-gating is applied.
6. Calculate time-gated signal strength and assign backscatter value to location \( r_0 \).

As stated earlier, the beamformer is designed to spatially focus the backscattered signals, compensate for frequency-dependent propagation effects, and optimally dis-
criminate against artifacts and noise. This is achieved through a frequency dependent 
filter that solves a penalized least squares problem designed to pass signals from a 
focused location in the breast with unit gain, subject to soft constraints on the norm 
of each filter [7].

DFT, Time-Aligning, and Windowing

The DFT of the received signal after the early-time artifact removal process is shown 
below:

$$X_i[l] = I(\omega_l)S_{ii}(r_0,\omega_l), \quad 1 \leq i \leq M, 1 \leq l \leq N$$

(5.1)

where $\omega_l$ is the frequency corresponding to the $l^{th}$ DFT index, $I(\omega_l)$ is the DFT of 
the transmitted pulse, and $S_{ii}(r_0,\omega_l)$ is an analytic model of the monostatic frequency 
response associated with propagation through breast tissue from the $i^{th}$ antenna to 
the scatter located at $r_0$ and back.

The time-aligned signals are windowed prior to the filtering stage to remove in-
terference and clutter that is present prior to time $n_a$:

$$g[n] = \begin{cases} 1 & n \geq n_a \\ 0 & otherwise \end{cases}$$

(5.2)

where $n_a$ is reference sample interval to which all received signals are time-aligned.

42
Defining \( \tau_i(r_0) \) as the round-trip propagation delay for location \( r_0 \) and the \( i \)th channel, \( n_a \) is chosen to be the worst case delay over all channels and locations:

\[
n_a = \max_{i,r_0} \tau_i(r_0)
\] (5.3)

**Minimum Beamformer Coefficients**

In order to design an efficient beamformer, it is desirable to use the minimum number of design frequencies and corresponding coefficients. This is determined in the following manner. The distance from the array to the furthest point in the scan region will determine the time extent for any received signal of interest, and consequently determines the minimum DFT length, \( N \). Along with the sampling frequency, \( \omega_s \), this determines the spacing between the DFT frequencies \( \Delta \omega \). Therefore, if the bandwidth of interest is \( B \), then the minimum number of coefficients, \( L \), is given by:

\[
L = \left\lceil \frac{B}{\Delta \omega} \right\rceil = \left\lceil \frac{BN}{\omega} \right\rceil
\] (5.4)

**Design Constraints**

Combining the phase delays associated with propagation and time alignment, the design constraint on \( W_i[l] \) can be written:
\[ I(\omega_l) \sum_{i=1}^{M} \tilde{S}_{ii}(r_0, \omega_l) W_i^*[l] = e^{-j\omega_l \tau_0 T_s}, \quad l_0 \leq l \leq l_0 + L - 1 \quad (5.5) \]

Stacking \( W_i \) and \( \tilde{S}_{ii} \) into the Mx1 vectors \( \mathbf{W}[l] \) and \( \mathbf{\tilde{S}} \) and rewriting in compact form we have:

\[ \mathbf{W}^H[l] \mathbf{\tilde{S}}(r_0, \omega_l) = e^{-j\omega_l \tau_0 T_s}, \quad l_0 \leq l \leq l_0 + L - 1 \quad (5.6) \]

If this is satisfied, then the beamformer output will be:

\[ Z(\omega_l) = e^{-j\omega_l (\tau_0 + n_a) T_s} \quad (5.7) \]

After converting back to the time-domain and windowing, the output energy, \( p(r_0) \), can be found by taking the sum of the squares of the time-gated signal:

\[ p(r_0) = \sum_n |z[n]h[r_0, n]|^2 \quad (5.8) \]

**Filter Weight Design**

Equation 5.6 has a closed form exact solution given by:
\[
W[l] = \frac{\hat{S}(r_0, \omega_l)e^{-j\omega_l r_0 T_s}}{\hat{S}^H(r_0, \omega_l)\hat{S}(r_0, \omega_l)} \tag{5.9}
\]

Here we see one of the main advantages of solving in the frequency domain: unlike the time domain solution, no matrix inversion is required to determine the exact solution to \(W[l]\). However, the exact solution is not robust since the magnitude of the weights can become arbitrarily large. This is especially problematic under conditions where the signal strength in the propagation model is small. As shown below, the robustness of the beamformer is proportional to the norm of the weight vector or noise gain:

\[
G_{ls}[l] = \mathbf{W}^H[l] \mathbf{W}[l] = \frac{1}{\hat{S}^H(r_0, \omega_l)\hat{S}(r_0, \omega_l)} = \frac{1}{|I(\omega_l)|^2 \sum_{j=1}^{M} |S_{jj}(r_0, \omega_l)|^2} \tag{5.10}
\]

Instead, a penalized least squares solution is used, which balances the norm of the coefficient against the approximation error. Using this method, the robustness is given by:

\[
G_{pls}[l] = \mathbf{W}^H[l] \mathbf{W}[l] = \frac{M}{1 + |I(\omega_l)|^2 \sum_{j=1}^{M} |S_{jj}(r_0, \omega_l)|^2} \tag{5.11}
\]
In this way, the noise gain cannot exceed M, and the robustness of the beamformer has been improved.

**Window Design**

If the beamformer satisfies equation 5.6 and the tumor is assumed to be a point scatterer, then the output is a bandlimited impulse response that has been time-shifted, attenuated and sampled. This is a good assumption since the goal of microwave imaging is to detect very small tumors. Since both the bandwidth and the time shift are known, the location and width of the main lobe containing the majority of the backscattered energy are both known. A natural choice for the windowing function is to pass only the time samples corresponding to the main lobe:

\[
h[r_0, n] = \begin{cases} 
1 & n_h \leq n \leq n_h + l_h \\
0 & \text{otherwise}
\end{cases}
\]  

(5.12)

### 5.1.3 Hypothesis Testing

Once the backscattered energy has been found for each voxel, the MIST system must identify and locate the positions of suspected tumors. This hypothesis testing is done using a generalized likelihood ratio test (GLRT). In the GLRT, the backscatter signal, \(y_i\), is assumed to contain only signal, \(\alpha_{t0}s_i(r_{t0})\), clutter \(c_i\), and noise \(n_i\):
\[ y_i = \alpha_{l_0}s_i(r_{l_0}) + c_i + n_i \]  \hspace{1cm} (5.13)

The signal is assumed to be known. If no scatterers are present (null hypothesis), then the scale factor \( \alpha_{l_0} \) is zero, otherwise (alternative hypothesis) it has a nonzero value that is deterministic and unknown. The clutter and noise are assumed to be zero-mean Gaussian distributed. The covariance structure \( \mathbf{R} \) is assumed to be known, but the power level, \( \sigma^2 \) is unknown. Under these assumptions, the GLRT test statistic is given by the likelihood ratio of the two hypotheses using the maximum likelihood estimates of the unknown parameters \( \sigma^2 \) and \( \alpha \):

\[
\Lambda_l = \frac{p(x; A_l, \hat{\alpha}, \hat{\sigma}^2)}{p(x; H_l, \hat{\alpha}, \sigma^2)}
\]  \hspace{1cm} (5.14)

For the given data model, this reduces to the ratio of the unbiased variance estimates under the null and alternative hypotheses raised to the NM/2 power:

\[
\left( \frac{\hat{\sigma}^2|H_l}{\sigma^2|A_l} \right)^{NM/2}
\]  \hspace{1cm} (5.15)

When the test statistic exceeds the threshold, the null hypothesis is rejected, and a scatterer is detected at location \( r_{l_i} \).[36]

47
5.2 TSAR Imaging System

The TSAR imaging method can be divided into two main stages: a preprocessing stage that is designed to eliminate the large backscatter response of the skin and a beamforming stage that is a modified version of the confocal beamformer.

Skin Subtraction and Clutter Reduction

The skin subtraction algorithm used in the TSAR imaging system is based on the recursive least squares (RLS) algorithm, an adaptive filtering method. The main difference between the RLS algorithm and the MIST skin subtraction approach is that in RLS, the weight vectors are updated after each time step, while in MIST, a constant weight vector is shifted through the selected window. In TSAR, the RLS algorithm is applied to the time steps corresponding to points outside the breast up until the interior surface of the skin, while an adaptive correlation method called Woody Averaging is applied to the remainder of the signal. The two estimated signals are then combined and subtracted from the target signal [37].

Confocal Beamforming with Improved Clutter Reduction

The beamformer in the TSAR system is a slightly modified version of the confocal beamformer intended to offer improved clutter reduction. The output of the TSAR
beamformer (backscatter strength at a point in space) can be described by the following equation [37]:

\[
p(i, j, k) = \left( \sum_n s(i, j, k, n)w(i, j, k, n)Q(i, j, k, n) \right)^2
\]

(5.16)

As can be seen from the equation, the backscatter strength at any point in space is a function of each antenna, \(s(i, j, k, n)\), a weighting function, \(w(i, j, k, n)\), and a compensation factor, \(Q(i, j, k, n)\). In TSAR, the weighting function gives more weight to antennas that are closer to the focal point. The signal compensation function compensates for the different properties of breast tissue, skin, and the surrounding fluid. While the TSAR beamformer lacks the complexity of the MIST system, it offers a number of innovative features and, as will be shown later, produces fairly competitive imaging results.
Chapter 6

Tomographic Imaging Systems

In tomographic microwave imaging systems, a set of measurement data is collected using antennas on the surface of a chamber, and the unknown complex permittivity distribution of the material inside can then be found using one of a variety of methods that have been developed to address this ill-posed inverse problem. These systems can generally be categorized into two main types based on whether the inverse problem is solved for a series of two dimensional slices using data collected from a planar array of antennas (2D imaging) or for the entire volume using data collected from antennas positioned throughout the perimeter of the chamber (3D imaging).

6.1 2D Imaging Systems

The imaging system developed by Meaney et al. [25] represents the prototypical 2D microwave imaging system and is the only MWI system thus far to be used in clinical trials. In the initial prototype system developed at Dartmouth College in 2002, the imaging data was obtained by a set of 16 transmit/receive monopole antennas that operated from 300 MHz. - 1 GHz. Data was acquired by illuminating the volume
sequentially at 12 discrete frequencies with each individual antenna and measuring
the scattered signal using all other antennas in parallel. A computer controlled
translating motor was used to step the array through 7 vertical positions. A saline
solution was employed as a coupling medium despite its poor match to healthy breast
tissue. Inversion was done using an algorithm based on the iterative Gauss-Newton
method. Using this system, saline anomalies inside excised breast tissue as small as
1.1 cm were successfully imaged, and objects as small as 4mm were detectable [31].

The current Dartmouth system [25] offers a number of improvements to the old
system. An array of 16 Monopole antennas stepped through 7 vertical positions
is still used, but the frequency range has been increased to 500 MHz - 2.3 GHz.,
which allows for increased resolution. In addition to increasing the frequency, the new
system features a number of improvements to the source, switching, and measurement
electronics. Also, a glycerine/water mixture that is less conductive and is better
matched to the known properties of breast tissue has replaced saline as the coupling
medium. Finally, the iterative Gauss-Newton method used for inversion has also been
improved. Details of this improved system are provided in the following sections.
6.1.1 Data Collection Methods and Hardware System Design

The hardware system developed by the Dartmouth group is the result of design philosophy that emphasized simplicity, reliability, and modularity. All of these are desirable qualities in a prototype system but can require tradeoffs in measurement speed and spatial resolution. Figure 6.1 (a) shows the modular nature of the Dartmouth imaging system, made up of six distinct hardware components: an RF source, power divider network, switching matrix, transceivers, imaging chamber, and measurement
electronics.

**Superheterodyne RF Source and Measurement Electronics**

An Agilent ESG RF signal generator produces the single-sideband, carrier-suppressed modulated signal that is used to illuminate the chamber, the coherent carrier reference signal that is used to demodulate the received signal, and the I/Q signal outputs needed for I and Q recovery with the lock-in amplifier. Measurement of the demodulated IF signal is done in parallel using a 333kHz data acquisition board (NI 6025E) after a preamplification stage (NI SCXI-1125). After the signals have been digitized, the final stage of recovering the amplitude and phase components of each signal is done using a software based lock-in amplifier [8].

**Switching and Power Divider Networks**

As a superheterodyne system, both the modulated signal used to illuminate the chamber by a transmitting antenna and the coherent LO signal used to recover the IF signal must be routed to each transceiver. In order to accomplish this, an RF 1:32 switching matrix is used to route the modulated transmit signal and a corporate power divider network is used to supply the coherent reference signal to each transceiver. The RF switching matrix is composed of 3 levels of switches that intro-
duce approximately 3 dB of insertion loss and provide roughly 40 dB of interchannel isolation. Since this level of isolation is insufficient, additional switching and amplification circuitry is introduced in the individual transceiver modules. For the power divider network, a five-level network of two way power dividers was used. However, in order to compensate for the high levels of insertion loss and mismatch associated with the power dividers, a combination of buffer amplifiers and matched attenuators was introduced to address these issues [8].

Transceiver Module

In the Dartmouth system, each antenna has a dedicated transceiver module (shown in Figure 6.1 (b)) which allows for parallel data acquisition and dual antenna mode operation—antennas are electronically switched between transmit and receive mode using a set of computer controlled switches. In addition to selecting the operating mode of the antenna, these switches introduce an additional 60 dB of isolation between channels. Furthermore, giving each antenna its own buffer amplifier in transmit mode and low noise amplifier (LNA) in receive mode increases system SNR and serves to help isolate the signal on each channel from coupling back through the switching network to other channels. Finally, the balanced mixer is able to convert the received
RF signal down to a frequency that can be measured with the 16 bit parallel A/D board [8].

**Matching Fluid and Imaging Chamber**

![Graph showing varying permittivity and conductivity of matching fluids](image)

**Figure 6.2** Dartmouth Matching Fluid Formulations [8].

A mixture of glycerin and water is used as a coupling medium since it provides a good match to the permittivity of healthy breast tissue in the frequency range of interest, is well-tolerated for skin contact, and provides a sterile environment. Since the permittivity of healthy mammary tissue varies with the fat content of the breast, it is possible to adjust the exact permittivity of the fluid to improve matching by varying the ratio of glycerin to water as shown in Figure 6.2. As can be seen in the same figure, the conductivity of the fluid is relatively high, which is a limiting factor in both the size of the effective imaging domain and the spatial resolution of the
The imaging chamber consists of the 16 element monopole antenna array which is connected using a set of rigid coaxial cables that slide through Teflon seals located at the bottom of the chamber. The vertical position of the array is set using a computer-controlled linear actuator and stepper motor capable of a resolution of 0.05 mm. The coupling fluid is pumped from a reservoir into the chamber, and the entire system is integrated into an examination table where the patient lies prone during the exam [8].

**System Characterization**

The Dartmouth imaging system boasts an impressive noise floor of -108 dBm and dynamic range of 120dB. The low noise figure is due in large part to their choice of low noise amplifier in the receiver channels. The equally impressive dynamic range figure is primarily achieved by the incorporation of variable gain settings from 1 to 2000 for each receiver channel and by intelligently varying the output power to prevent saturation [8].

Channel cross talk was measured at 15 dB above the noise floor for the channels adjacent to the selected receiver and indistinguishable from noise for all other receiver
channels. The authors argue that since the adjacent channels contain similar information (field magnitude) to the selected channel, this slight lack of isolation should have little effect on the selected signal. In either case, through the intelligent use of layered switches, the system has achieved an impressive level of interchannel isolation [8].

The quoted acquisition time for a set of 2D slice data at a single frequency (16 transmitters x 15 receivers) is 8 seconds [8]. Presumably, a full set of data for a single breast image (slice data at 12 discrete frequencies for each of 7 vertical positions) takes a minimum of 11 minutes, neglecting the time needed to transition to the next frequency and reposition the array at each vertical position. While the long data acquisition time is perhaps the greatest disadvantage of this system, this is mitigated by the fact that patients undergoing a microwave imaging procedure suffer significantly less discomfort than they would in the case of a mammogram.

6.1.2 Forward Modeling and Inversion Methods

The foundation of the Dartmouth group’s inversion algorithm is a Gauss-Newton iterative method that solves the minimization problem:
\[ \vec{k}^2 = \arg \min \left\{ \| \vec{E}_{l-p}^{\text{meas}} - \vec{E}_{l-p}^{\text{calc}}(\vec{k}_n^2) \|_2 \right\} \quad (6.1) \]

where \( \vec{E}_{l-p}^{\text{meas}} \) is the measured complex scattered electric field data given by

\[
\vec{E}_{l-p}^{\text{meas}} = \begin{bmatrix}
\log |\vec{E}_{\text{target}}^{\text{meas}}| - \log |\vec{E}_{\text{empty}}^{\text{meas}}| \\
\text{phase} \vec{E}_{\text{target}}^{\text{meas}} - \text{phase} \vec{E}_{\text{empty}}^{\text{meas}}
\end{bmatrix}
\quad (6.2)
\]

and \( \vec{E}_{l-p}^{\text{calc}}(\vec{k}^2) \) is the calculated complex scattered electric field data from the forward model given by:

\[
\vec{E}_{l-p}^{\text{calc}}(\vec{k}^2) = \begin{bmatrix}
\log |\vec{E}_{\text{target}}^{\text{calc}}(\vec{k}^2)| - \log |\vec{E}_{\text{empty}}^{\text{calc}}(\vec{k}_o^2)| \\
\text{phase} \vec{E}_{\text{target}}^{\text{calc}}(\vec{k}^2) - \text{phase} \vec{E}_{\text{empty}}^{\text{calc}}(\vec{k}_o^2)
\end{bmatrix}
\quad (6.3)
\]

and \( \vec{k}^2 \) is the complex wave number squared:

\[ \vec{k}^2 = \omega^2 \mu_0 \epsilon + j \omega \mu_0 \sigma \quad (6.4) \]

**Gauss-Newton Algorithm**

The Gauss-Newton algorithm can be broken down into the following steps [25]:

1. Call forward solver to calculate the electric fields from the current constitutive parameters \( \vec{k}_n^2 \) and to check for convergence.
2. Calculate the Jacobian $\bar{J}$ for $\bar{k}_n^2$.

3. Calculate the updated values of the constitutive parameters $\bar{k}_{n+1}^2$.

Since the forward solver step is far more computationally expensive than the latter step of updating the constitutive parameters, the overall efficiency of the algorithm is governed by the speed of the forward solver and by the number of iterations needed before the convergence condition is met. Furthermore, the number of iterations is largely determined by the accuracy of the constitutive parameter update. Therefore, it is useful to examine these two critical steps in more detail.

**Forward Solver**

The forward solver used by the Dartmouth group is a hybrid algorithm that uses finite elements to represent the heterogeneous scattering problem within the imaging domain and boundary elements to represent the homogeneous space outside the imaging domain. The effect of the antennas not currently acting as an active transmitter or receiver is accounted for by representing them as electromagnetic sinks located in the boundary element zone [38, 39]. The efficiency and accuracy of this solver arises from its hybrid approach: the boundary elements eliminate the need for approximate boundary conditions and reduce the need for finite-elements to the imaging domain.
Tikhonov Regularization: The GN-T Method

The original method that the Dartmouth group used to compute the updated constitutive parameters was based on a Tikhonov regularization procedure that consists of three steps:

1. Obtain the Newton direction $\bar{d}_n$ by solving the normal equation of

$$\bar{J}(k_n^2)\bar{d}_n = \bar{E}_{l-p}^{meas} - \bar{E}_{l-p}^{calc} (k_n^2)$$

(6.5)

which can be written (suppressing the argument of the Jacobian):

$$\bar{J}^T \bar{J} \bar{d}_n = \bar{J}^T (\bar{E}_{l-p}^{meas} - \bar{E}_{l-p}^{calc} (k_n^2))$$

(6.6)

Since the above equation is ill-conditioned, regularization must be performed and in this case, the Tikhonov algorithm is used, resulting in the following linear equation which can be solved for $\bar{d}_n$:
\[
\left( \bar{J}^T \bar{J} + \lambda \bar{I} \right) \bar{d}_n = \bar{J}^T \left( \bar{E}_{\text{meas}} - \bar{E}_{\text{calc}}(\bar{k}^2_n) \right)
\]  

(6.7)

2. Determine the Newton step \( \alpha_n \).

While algorithms exist for determining the optimum value of \( \alpha_n \), they all require multiple iterations of the forward model to compute, so the Dartmouth group has implemented a simplified method that determines the Newton step based solely on the iteration number, starting with 0.1 for the first three iterations and gradually increasing to 0.5 after 12 iterations. This choice of \( \alpha_n \) is designed to keep the step size small enough to ensure slow changes in the phase distribution of the computed fields and maintain the stability of the algorithm.

3. Having computed \( \bar{d}_n \) and \( \alpha_n \), the values of the constitutive parameters can then be updated using:

\[
\bar{k}^2_{n+1} = \bar{k}^2_n + \alpha_n \bar{d}_n
\]

(6.8)

This completes the update stage of the GN-T algorithm, and the iterative process repeats as the forward solver is once again called to compute the fields and check for convergence.
Conjugate Gradient Optimization: The GN-C Method

In the current version of the Dartmouth system, the Tikhonov algorithm has been replaced with a conjugate gradient least squares (CGLS) method that reduces the number of iterations required to reach convergence and appears to offer improved imaging results. The CGLS method is an iterative approach that is able to work directly on the Jacobian matrix instead of having to compute $J^T J$. It should be noted that this approach allows greater control over the regularization effects since they are a function of how many iterations the algorithm is allowed to run. Adjusting the number of iterations also controls the balance between spatial resolution and image artifacts (more iterations result in increased resolution but additional artifacts). Unfortunately, further details of the CGLS method are beyond the scope of this document; however, interested readers are referred to [25] for specific details of its application to the Dartmouth system.

6.2 3D Imaging Systems

3D MWI systems, such as those being developed at Duke, have the potential to offer improved resolution and data collection efficiency over their 2D counterparts.
6.2.1 Data Collection and System Design

There are two possible approaches to data collection in a 3D microwave imaging system and each has its own advantages. The first method is to employ a set of CNC (computer numerical control) positioners to independently scan a pair of antennas throughout the perimeter of the imaging volume while a set of transmission measurements are made. This method offers a number of advantages similar to the 2D imaging system described in the previous section. First, since there are only two antennas, each antenna is directly connected to its corresponding port (transmit/receive) on the measurement device without the need for RF switching network. This offers the obvious benefit of increasing SNR, as well avoiding all of the other disadvantages associated with electronic switching systems, such as cross channel coupling. In addition, mutual coupling effects caused by nearby antennas are avoided. In general, this method greatly simplifies the system design, forward modeling, and image inversion process and offers greater measurement flexibility at the cost of greatly increased measurement time, possibly to the detriment of SNR and spatial resolution.

The other method involves using a fixed 3D array of antennas positioned throughout the surface of the imaging chamber and connected to the source and measurement
devices through an RF switching network. Advantages of this method include faster data collection, more accurate antenna positioning, and integration of the antenna array with the imaging chamber. Further details of this approach will be given in subsequent chapters.

### 6.2.2 Forward Modeling

Computing electromagnetic fields given the properties of the medium and its geometry is the forward modeling process. This forward problem is essential for the system characterization and calibration, and for the nonlinear image reconstruction (iterative inverse scattering method).

Forward modeling for microwave imaging involves the numerical solution of Maxwell’s equations, as analytical solutions are in general not possible for an inhomogeneous
Numerical solution methods for such problems can be categorized as time-domain and frequency-domain methods. Since the measurements in tomographic microwave imaging systems are performed at discrete frequencies, it is generally more efficient to solve the problem by using frequency-domain methods. At Duke, a fast volume integral equation method is used rather than a finite-element method or a hybrid finite-element/boundary-element method in the frequency domain.

**Fast Forward Solvers**

The conventional approach for solving volume integral equations in the frequency domain is the method of moments (MOM) [40]. However, given the complex nature and large number of unknowns, using the MOM for volume integral equations is too computationally expensive. For example, at 6 GHz for high-frequency microwave imaging, for an accurate discretization of a voxel size of 1.25 mm, a volume of size $16\times16\times16\text{ cm}^3$ in MWI breast imaging represents 6.3 million unknowns in the integral equation.

To address this issue, researchers at Duke have been developing more computationally efficient and less memory-intensive forward solvers. The first of these is a more efficient approximate scattering method, the diagonal tensor approximation...
(DTA), which significantly accelerates the forward solution of the volume integral equations. In simple terms, this is because using the FFT in the DTA decreases the computational cost of the conventional extended Born approximation method from $O(N^2)$ to $O(N \log N)$ arithmetic operations. Also, because it is source-dependent, the DTA has the potential to provide a better approximation than the existing extended Born approximation [41, 42]. In addition to the DTA, the Duke research group has made significant contributions to the development of full-wave fast forward solvers with their stabilized biconjugate-gradient fast Fourier transform (BCGS-FFT) method, which has been shown to be several times faster than the CG-FFT method for typical applications [43, 44, 45]. Clearly, without the development of these state-of-the-art fast forward solvers, full 3D microwave imaging would not be possible.

**Commercial Solvers**

While commercial solvers such as Ansoft’s HFSS and CST’s Microwave Studio are too inefficient to be used as forward solvers in an inversion process, they can still play an important role in the forward modeling of the microwave imaging system. First and foremost, thanks to their powerful graphical user interfaces, optimization routines, and reporting tools, they are excellent tools for the design of all aspects
of the imaging system (e.g. antenna arrays, switching system, matching fluid, and imaging chamber). In addition to their usefulness in the design stage, these commercial packages can also provide characterization of certain inherent aspects of the imaging chamber that can be incorporated into the fast forward solver and inversion algorithm to improve their accuracy. The first of these is the polarization of the transmit/receive antennas. Knowledge of the antenna polarization is critical in determining which components of the electric field are being measured on the perimeter of the chamber, which in turn are used by the inversion algorithm to solve for the electrical properties of the volume. Secondly, models of the imaging chamber using these solvers can be used to generate a Green’s function that describes the full response of the inhomogeneous space that is the imaging chamber filled with matching fluid. This Green’s function characterization of the imaging chamber can then be incorporated into the fast solvers that do not include a full model of the complex antenna geometry. By using this hybrid approach, the efficiency of the fast solvers can be exploited without sacrificing the accuracy of the full solvers.
6.2.3 Inversion Methods

In 3D microwave imaging, it is necessary to solve a nonlinear inverse scattering problem that can be described by the following volume integral equation:

\[
E(r) - E^{inc} = \omega^2 \mu_b \left(1 + \frac{\nabla \nabla \cdot k_b^2}{k_b^2}\right) \int_V E(r') g(r, r') [\epsilon^*(r') - \epsilon_b^*] dV' \tag{6.9}
\]

where \(\mu_b\) and \(\epsilon_b^*\) are the permeability and complex permittivity of the homogeneous background medium, while \(\epsilon^*\) is the complex permittivity of the imaged domain. It is assumed that the medium is non-magnetic, thus \(\mu = \mu_b\) in the imaged domain.

At Duke, the distorted Born iterative method and the contrast source inversion method are used to solve the inverse problem [32, 33, 34, 46]. To reconstruct the unknown profile of dielectric constant and conductivity from \(M\) microwave measurements, the volume is discretized into \(N\) voxels, each with an independent value of dielectric constant and conductivity. Next, one of two inverse scattering approaches are taken:

1. A nonlinear inverse method based on the diagonal tensor approximation and Born iterative method (BIM).

2. The DTA preconditioned contrast source inversion (CSI) method or distorted Born
iterative method (DBIM).

In the first method, the nonlinear problem is solved iteratively by the Born iterative method and DTA. Within each iteration of the BIM, the contrast function is updated by the DTA through a regularized least-square optimization. Since the volume integral equation is solved approximately by the DTA, this first step is usually much faster than the second step. For a problem with $N$ unknown voxels and $M$ measurement data points, the computational time of this inverse DTA procedure is reduced to $O(M_T N \log N)$. Furthermore, the memory requirement is reduced to $O(M_T N)$ ($M_T$ is the number of transmitters) [46, 34].

In the second inverse method, the contrast source inversion (CSI) method or the distorted Born iterative method (DBIM) are used, but the DTA inversion result is used as the initial solution in order to accelerate convergence. Finally, in CSI and DBIM with the full-wave BCGS-FFT method, a super-resolution better than a quarter wavelength has been achieved [34, 47, 9].
Chapter 7

Antenna Design for Diagnostic Breast Imaging

7.1 General Design Considerations

Imaging human tissue with microwaves introduces a number of unique design challenges. Foremost among them is the fact that the vast majority of antenna theory assumes that the antenna is radiating into free space. For near field breast imaging, this is not the case. Instead, antennas must be designed to radiate either directly into tissue, or into a matching fluid media designed to mimic the dielectric properties of the tissue to be imaged. This is because, when waves traveling in free space encounter human tissue, the large dielectric mismatch causes significant reflections, making imaging inside the tissue extremely difficult. Therefore, when designing antennas for microwave breast imaging, all of the preexisting design equations and theory developed for free space radiation must be modified to account for radiation directly into higher dielectric media.

In addition, several other design factors must be considered when developing an antenna to be used in a breast imaging system. Perhaps the most important of
these is sampling resolution. Since the antennas are to provide a description of the field distribution at the perimeter of the imaging volume that will then be used to reconstruct the dielectric properties of the material inside, the antenna design should seek to maximize the number of antennas that can be packed on the surface of the volume. A larger number of antennas on the surface increases the sampling density of the field distribution, thus theoretically improving the quality of the reconstructed image.

Unfortunately, the need for a high density of antennas is in direct conflict with another primary design consideration: the need to minimize mutual coupling between antennas. Mutual coupling (or mutual impedance) between nearby antennas is the result of the current flowing in one antenna inducing current in other nearby antennas. This induced current could then reradiate, possibly returning a significant amount of energy to the source antenna. Nearby antennas could also alter the shape of the field lines surrounding the source antenna (typically by providing additional ground), thus altering the radiation properties of the antenna. All of these mutual impedance effects can negatively impact antenna performance in a variety of ways (shift resonant frequency, alter radiation pattern, reduce matching performance). Furthermore,
these effects can influence each antenna in an array differently, depending on the geometry of the antenna and the array. Finally, for complex antenna geometries or array configurations, it is not always feasible to analytically compute exact mutual impedance effects. Therefore, it is important not only to design both the individual antenna and the array to minimize these mutual coupling effects, but also to perform simulations and measurements on prototype arrays to ensure that mutual coupling will not affect antenna performance.

The final design consideration of importance is a practical one: ease of rapid, precise, reliable, and low cost fabrication. While a theoretical antenna design may meet all of the above requirements and have many other desirable qualities, if antenna arrays cannot be made in an efficient, reliable, and precise manner, then the antenna design is of little use for microwave imaging applications.

Since microwave imaging is done at discrete frequencies, fabrication tolerance requirements can be thought of as a function of antenna bandwidth. While all of the antennas in the system must operate efficiently at a single common frequency, finding that frequency is made easier by the fact that each antenna has a range of frequencies that lie within its operating bandwidth. Therefore, fabrication methods need only be
precise enough so that the resonant frequency of each antenna lies within a range of
frequencies determined by the bandwidth of the antennas. In addition, the fabricated
antenna array should be able to reliably tolerate exposure to a variety of matching
fluid materials and should be easily arranged into an imaging chamber configuration
when combined with other arrays. Finally, the ideal antenna array should be low cost
and have a short manufacturing time (preferably using readily available materials
such as PCB etching on FR-4 substrate).

While there are many different antenna types that could potentially be used for
microwave imaging, some are more naturally suited to meet the above requirements
than others. As discussed in the following section, printed circuit antennas, such
as the microstrip patch antenna, are particularly well suited for use in microwave
imaging systems.

7.2 Microstrip Antenna Theory: Analysis and Design

While the first printed antennas were proposed by Deschamps in 1953 [48], it wasn’t
until the early 1970s that the first practical antennas would be developed by Howell
[49] and Munson [50]. This was mainly due to advancements in substrate material
properties and photo-etching techniques, which, coupled with better theoretical models, enabled the field of printed antennas to blossom. Since that time, a variety of printed antennas have come into use (such as the printed dipole and the stripline slot antenna), but perhaps the most prominent of these is the microstrip antenna [51].

![Figure 7.1 Patch Antenna with Microstrip Feed.](image)

Microstrip antennas can take many forms, but in general, they consist of a radiating copper patch on a dielectric substrate with a ground plane on the opposite side (see Figure 7.1). The radiating patch can take many shapes, with the rectangular patch and the circular patch being among the most common. Patch antennas are most commonly fed with a microstrip transmission line or coaxially with the center conductor of the connector fed through the substrate and connected to the patch.
7.2.1 Radiation Mechanism

A logical place to begin an analysis of any antenna is to examine the method by which it radiates. In the case of microstrip patch antennas, radiation occurs from the fringing fields that exist between the edge of the conducting patch and the ground plane beneath it. In a simple case, such as a rectangular patch antenna roughly $\frac{\lambda}{2}$ in length that is coaxially fed, radiation will occur at the open-circuited ends of the patch (see Figure 7.2). Since the patch is a half-wavelength long, the components of the electric field normal to the patch will be out of phase, while the tangential components will be in phase (see Figure 7.3).

![Figure 7.2](image)

**Figure 7.2** Radiation Due to Fringing Fields From “Virtual Apertures” (in Red) on the Ends of the Patch.

In terms of radiation, the in-phase tangential components will add and give strong radiation normal to the patch. Depending on patch geometry and the size of the
Figure 7.3 Field Distribution Along Length of Rectangular Patch.

ground plane, the normal fields may also add to yield strong radiation along the patch (as seen in Figure 7.4). As a result, the E-plane of the antenna (the plane parallel to the E-field vectors containing the strongest radiation) bisects the antenna symmetrically through the feed point, while the H-plane (plane parallel to the H-field containing the strongest radiation) also bisects the antenna symmetrically, but does not contain the feed (since the feed is offset).

Finally, since the fields add in this manner, the polarization of the far-fields is linear, pointing parallel to the patch for fields radiating in the broadside (normal) direction, and the linear polarization varies as the tangent of a circle surrounding the patch until finally the fields are linearly polarized normal to the patch for fields propagating in the endfire (transverse) direction. For further discussion of the radiation characteristics of patch antennas, the reader is referred to [52, 53, 54, 55].
7.2.2 Theoretical Models of Analysis

Many theoretical models have been developed that attempt to capture the physics of microstrip antenna operation, and thus enable the antenna designer to analytically predict and control the performance of microstrip radiators. As yet, no exact model exists, and even if such an exact solution did exist, it would likely not take the place of the many approximate models that have been developed to suit a variety of situations. Some of these models are extremely complex (but potentially more accurate), while others are relatively simple, but offer valuable insight from a design perspective. Excellent summaries of many of these methods are given in [51] and [56], and more detailed discussions of the individual models can be found in [57, 58,
59, 60, 61, 62, 63, 64, 65, 66, 50, 67]. Of these models, a brief description of two of the more complex models, the Vector Potential and Green’s Function models, will be given here, and a slightly more in-depth discussion of the simpler transmission line model (used in the design of the antennas in this project) will follow.

**Vector Potential Approach**

In the vector potential solution, the microstrip patch is treated as a horizontal electric dipole pointing in the $\hat{x}$ direction on the surface of a dielectric slab. The vector potentials due to a horizontal dipole on a dielectric slab of thickness $h$ over a ground plane ($\hat{z}$ normal) are then determined by applying the boundary conditions at the air-dielectric interface that require the vector potential $\vec{A}$ to have components in the $x$ and $z$ directions that satisfy the inhomogeneous Helmholtz equation:

$$\left(\nabla^2 + k^2\right)\vec{A} = -\mu \vec{J} = -\hat{x} \mu Idx\delta\vec{r} \tag{7.1}$$

From the vector potential, the fields can be recovered using the well known equations:

$$\vec{H} = -\frac{1}{\mu} \nabla \times \vec{A} \tag{7.2}$$
\[ \vec{E} = \frac{1}{j\omega \mu \varepsilon} [k^2 \vec{A} + \nabla (\nabla \cdot \vec{A})] \quad (7.3) \]

This leads to an integral equation for the fields which has no analytical solution and must be solved numerically. This makes the vector potential approach undesirable as a design tool, since it lacks a closed form solution and it is difficult to ascribe any clear physical meaning to the above equations [51].

**Green’s Function Approach**

Another approach to analyzing microstrip antennas is to employ the dyadic Green’s function method of analysis. For the case of microstrip antennas, an appropriate choice of Green’s function is the one developed by Alexopoulos et al [58] using a Hertzian dipole printed on a grounded substrate:

\[ G(r/r') = -\frac{30j}{\pi k_\alpha} \int_{-\infty}^{\infty} \int_{-\infty}^{\infty} \psi(k_x, k_y) e^{-j[k_z(x-x') + k_y(y-y')]} \frac{F_x F_z}{F_x F_z} dk_x dk_y \quad (7.4) \]

where
\[ \psi(k_x, k_y) = \]
\[
\left( (k_o^2 - k_x^2) \beta \tanh \beta h + (k_x^2 \varepsilon_r - k_o^2) \beta_o \right)\beta - k_x k_y (\beta_o + \beta \tanh \beta h) \\
- k_x k_y (\beta_o + \beta \tanh \beta h) \right) \left( (k_o^2 - k_y^2) \beta \tanh \beta h + (k_o^2 \varepsilon_r - k_y^2) \beta_o \right) \]

(7.5)

with \( F_x = \beta_o + \beta \coth \beta h \) and \( F_z = \beta_o \varepsilon_r + \beta \tanh \beta h \).

Using this Green’s function and taking the following integral of the current density over the surface of the patch,

\[ \tilde{I}(\theta, \phi) = \int \int_{S} \tilde{J}(r^{'}) e^{j k_o \sin \theta (x' \cos \phi + y' \sin \phi)} ds' \]

(7.6)

the fields due to the patch can then be computed (if the patch is assumed to be a two-dimensional transmission line in order to determine the appropriate boundary conditions for the surface current density). Like the vector potential approach, this is an extremely tedious process and does not naturally lead to a useful set of design equations.

**Transmission Line Model**

While both of the above approaches may be somewhat satisfying from a rigorous theoretical standpoint, both fail to offer a great deal of physical intuition into how patch antennas work, nor do they easily translate into a design methodology. On the other hand, the transmission line model, which was first developed by [50] and
[67], offers a straightforward analysis based on transmission line theory that naturally leads to a set of simple equations relating the geometry of a rectangular patch, its material properties, and the feed location to the predicted resonant frequency and input impedance.

The transmission line approach generally treats the microstrip patch as a line resonator with no transverse field variations. The fields vary along the length (typically \( \frac{\lambda}{2} \)) and radiation occurs mainly from the fringing fields at the open-circuited ends. The patch may then be represented as two slots spaced a distance \( L \) apart, and each slot is considered to radiate the same field as a magnetic dipole with magnetic current of:

\[
\vec{M} = \hat{z}2E_x = \frac{\hat{z}2V_o}{h}
\]

where the factor of 2 is due to the image caused by the ground plane.

From this, the fields can be computed in the usual manner as discussed in [51]. More importantly from a design perspective, however, this model also predicts (using transmission line theory) the input admittance of a rectangular patch antenna of length \( L \) and width \( W \) that is coaxially fed (feed centered along \( W \), offset along \( L \)).
to be:

$$Y_1 = Y_0 \left( \frac{Z_0 \cos \beta L_1 + jZ_w \sin \beta L_1}{Z_w \cos \beta L_1 + jZ_0 \sin \beta L_1} + \frac{Z_0 \cos \beta L_2 + jZ_w \sin \beta L_2}{Z_w \cos \beta L_2 + jZ_0 \sin \beta L_2} \right)$$  \hspace{1cm} (7.8)$$

where $L_1$ is the distance from the feed to the near edge, $L_2$ is the distance from
the feed to the far edge, $Z_0$ is the characteristic impedance of the microstrip and

$$Z_w = \frac{1}{G_w + jB_w}, \text{ with } G_w \text{ and } B_w \text{ empirically determined to be the following (assuming
the patch is radiating into air):}$$

$$G_w = \frac{0.0836W}{\lambda_o}, \quad B_w = 0.01668 \frac{\lambda_i W}{h} \epsilon_e$$

$$\epsilon_e = \frac{\epsilon_r + 1}{2} + \frac{\epsilon_r - 1}{2} \left( 1 + \frac{12h}{W} \right)^{-1/2}, \quad \frac{\Delta l}{h} = 0.412 \frac{(\epsilon_e + 0.3)(W/h + 2.64)}{(\epsilon_e - 0.268(W/h + 0.8)}$$

Using these equations, the input impedance of any rectangular patch geometry
may be directly calculated. This allows the designer to make an initial guess as to
the appropriate geometry for the patch to operate efficiently at a given frequency.
Furthermore, this set of equations also allows the designer to predict how the input
impedance will vary with each parameter. This is incredibly useful when combined
with simulation tools such as HFSS and CST. In this way, after an initial design has
been chosen and simulated, intelligent adjustments can then be made in an iterative way to the various parameters (either manually or automatically using optimization routines in the simulation package) until an optimal design is found.
Chapter 8

Microstrip Antennas for Diagnostic Breast Imaging

Microstrip antennas offer a number of advantages which make them ideally suited for use in a microwave imaging system. Their principle advantage lies in the fact that, because they are printed antennas, they can be manufactured quickly with a high level of precision and repeatability at a very low cost. In addition, multiple antennas can be easily integrated into an array on a single printed circuit board (PCB), which can be fabricated in a matter of hours using simple photo-etching techniques and widely available PCB tools and materials. Furthermore, because these antenna arrays are light weight, low profile, and planar, they can naturally be used to form the sides of the imaging chamber itself. Finally, since the antennas are fabricated on printed circuit boards, all of the additional microwave circuitry necessary for the microwave imaging system (transmission lines, RF switching, power division, etc.) can be integrated directly on the same board as the antennas themselves. For all of these reasons, microstrip antennas are a very attractive practical choice for use in a diagnostic microwave imaging system.
Beyond the practical reasons, microstrip antennas also offer a number of desirable technical qualities which make them an ideal candidate for microwave imaging. First, the feed line and impedance matching are integrated directly into the antenna design, obviating the need for additional circuitry. Also, polarization can be controlled simply by appropriately choosing the feed position, and both linear and circular polarizations are possible. Dual frequency operation is also possible in many designs. Finally, the radiation pattern of microstrip antennas is attractive for imaging since the antennas generally radiate into a half plane, with the largest amount of gain in the broadside direction (into the volume to be imaged).

Microstrip antennas are not without disadvantages, but many of them do not negatively affect their application in microwave imaging, and the others can be minimized with careful design. For instance, patch antennas typically have a very narrow bandwidth, but imaging is done at a single frequency, so this is not a major concern. Patch antennas also have poor power handling, but again this is not a concern since diagnostic imaging is a low power application. Poor endfire radiation performance (of most patch designs) is similarly not an issue since the volume to be imaged is always in the broadside direction for all antennas.
Two significant design challenges that do affect the performance of microstrip antennas used in medical imaging applications are mutual coupling and radiation efficiency. The first issue is mainly a matter of geometry: imaging applications require a dense packing of antennas to maximize the sampling of the field distribution, but traditional microstrip antenna design requires that the antenna size be on the order of a half-wavelength. Furthermore, in order to obtain maximum signal transmission, a common polarization between antennas is desirable. These two factors, a dense packing of antennas with a common polarization, will invariably lead to significant mutual coupling effects. The second issue is a result of the loading imposed on the antenna due to the fact that it is radiating into tissue or matching fluid, rather than air. In some sense this is helpful, since it increases the effective dielectric constant (and therefore reduces antenna size), but it also introduces significant challenges for achieving an impedance match and efficient radiation.

Two microstrip patch antenna designs have been developed that attempt to resolve these two main issues in different ways. The first is a relatively low frequency (900 MHz.) multi-layer folded patch antenna design which enables a significant reduction in the footprint of the antenna (and therefore increased density) while achieving
a good impedance match. The second antenna operates at a higher frequency and employs a bowtie shaped patch design whose elements were arranged in a staggered grid to achieve a maximum density of antennas, while maintaining a good impedance match and minimal mutual coupling.

8.1 Folded Patch Antenna

In our first prototype microwave imaging system, a set of densely packed, small form factor antennas operating near 900 MHz. was required. In terms of bandwidth, the only requirement was that the 3dB bandwidth of the antennas should be an order of magnitude greater than the center frequency production tolerance. This was to ensure that all antennas shared common operating frequencies.

To meet the above requirements, a folded patch antenna design was used. Figures 8.1 and 8.2 show the geometry of the five layer folded patch antenna used in our system.

![Figure 8.1 Side View of Folded Patch Antenna.](image-url)
Like the traditional patch antenna, the design equations require that the electrical length between apertures be 180 degrees. For impedance matching purposes, the feed point should be placed off center, such that the total electrical lengths traveled are approximately 45 degrees (short side) and 135 degrees (long side). One side then appears inductive at the input while the other appears capacitive. This results in the cancellation of the reactive components, leaving a purely real input impedance which can be matched to the characteristic impedance of the system (typically 50 Ω).

However, for the folded patch, these electrical lengths are distributed between many layers. Thus, for an n layer folded patch, the total width required is reduced by a factor of n. Using the transmission line model described above to predict the effect of variations of the patch geometry on the input impedance curve, the folded patch width, height, and feed location were then optimized in HFSS to obtain an impedance match near 900 MHz with the antenna radiating into matching fluid.
Figure 8.3 5x5 folded patch antenna array.

Five 18 cm x 18 cm square panels each containing 25 folded patch antennas were placed on five sides of a Plexiglas tank which held matching fluid and objects to be scanned. One array of 25 antennas which forms one side of the cube is shown in Figure 8.3.

Input Impedance and Mutual Coupling

Figure 8.4 shows a comparison of the simulated input impedance for a single antenna and for a panel of 25 antennas. These results show that the spacing between antennas is sufficient, such that each antenna in the array will perform in the same manner as it would in isolation.
Figure 8.4 Input Impedance Comparison - Single Antenna vs. Full Panel (HFSS Simulated Results)

Resonant Frequency and Bandwidth

Figure 8.5 shows a comparison of the measured return loss for a single antenna in a 25 antenna panel radiating into matching fluid with the simulated return loss of a single antenna in isolation as predicted by HFSS. This confirms that the antennas are functioning as designed and that mutual coupling is not a significant issue.

Polarization and Radiation Pattern

Figure 8.6 shows the polarization of the local fields surrounding a single folded patch antenna. The fields in the broadside direction are linearly polarized and directed
Figure 8.5 Comparison of Measured and Simulated Return Loss

Figure 8.6 Folded Patch Antenna: Field Polarization
along the length of the patch. This is in agreement with theory, as they point in the same direction as the tangential component of the fields in the apertures. Fields in the endfire direction are also linearly polarized and are directed perpendicular to the plane of the patch. This is also in agreement with theory, as their polarization matches that of the normal component of the fields in the apertures.

Figure 8.7 shows the gain of the antenna for the E-plane ($\phi = 0^\circ$) and the H-plane ($\phi = 90^\circ$), as predicted by HFSS. This pattern is in general agreement with theory, showing strong radiation in the broadside direction, with a null in the endfire direction in the E-plane. It should be noted that the gain values in the positive z direction are attenuated due to the loss in the matching fluid. Therefore, the maximum gain of
Figure 8.8 The magnitude of S21 for the transmission (a) between two opposing panels and (b) between front panel and bottom panel.

The antenna is actually somewhat higher than is shown in the plots, but the relative values of the gain as a function of radiation direction (i.e., the shape of the pattern) are accurate.

Chamber Transmission Measurements

Figure 8.8 (a) shows a comparison of the S21 (transmission) parameter for five antenna pairs containing antennas on directly opposing side panels. This plot shows good transmission between all five pairs of antennas, with a consistent peak occurring around 925 MHz. This suggests that each of the antennas is resonating at approximately the same frequency, and also indicates that it should be possible to complete the imaging scan at a single optimal frequency where all antenna pairs are performing
Figure 8.8 (b) shows the transmission between the front panel and the bottom panel. This plot also shows acceptable levels of transmission between all pairs of antennas, with the lowest S21 curve peaking around -50dB. This was for the case of the center antenna on the front panel and the farthest corner antenna on the bottom panel, which represents the minimum level of S21 performance for any pair of antennas. Since this is still greater than 30dB above the measured noise floor of the system, all antenna pairs provided an adequate level of signal transmission.

### 8.2 Bowtie Patch Antenna

In order to improve imaging resolution, another imaging applicator was needed that operated at a higher frequency (2.7 GHz). In addition, from an image reconstruction standpoint, it was also desirable to design an antenna that had a simpler geometry than our previous folded patch design. At first, this led to the investigation of using a simple rectangular patch, but a simple rectangular geometry introduced a number of undesirable consequences that became immediately apparent.

First, in order for a rectangular patch antenna on a standard substrate of FR-4 to be impedance matched at 2.7 GHz while radiating into human tissue or a matching fluid, the width of the patch must be made prohibitively thin. This results in a
number of problems including reduced efficiency (smaller radiation apertures), narrow bandwidth, poor broadside radiation, and reduced fabrication tolerances. In order to overcome those difficulties, the rectangular patch design was modified by introducing a linear taper along the length of the patch from the feed to each aperture. The tapered design allows the antenna to maintain its input impedance match while maintaining a reasonably sized radiation aperture, thus resolving the issues associated with a very thin antenna design.

The transmission line model described above was again used to predict the effect of variations of the bowtie patch geometry on the input impedance curve, and the bowtie geometry was optimized using HFSS and CST to obtain an impedance match near 2.7 GHz. The final geometry is shown in figures 8.9 and 8.10.

Four rectangular panels 10 cm in height, each containing 8 bowtie patch antennas, were then joined to form a prototype imaging chamber with a total of 32 antennas,
**Figure 8.10** Bowtie Patch Geometry: Side View.

![Bowtie Patch Geometry Diagram](image)

**Figure 8.11** Imaging applicator using 32 bowtie patch antennas operating at 2.7 GHz.

![Imaging Applicator](image)

as shown in Figure 8.11.

**Input Impedance and Mutual Coupling**

Since the virtual apertures of these antennas are located at the open-circuited ends of the bowtie patch, it was especially critical to maintain adequate spacing in the vertical direction near these radiation apertures. In order to accomplish this, a staggered arrangement of antennas was used for the antenna array.
Figure 8.12 Input Impedance Comparison - Single Antenna vs. Full Panel (CST Simulated Results)

Figure 8.12 shows a comparison of the simulated input impedance for a single antenna and for a panel of 8 antennas. These results show that the spacing and staggered arrangement of the antennas is sufficient to minimize mutual coupling such that it does not have an impact on the resonant frequency, bandwidth and radiation pattern of any of the antennas.

**Resonant Frequency and Bandwidth**

Figure 8.13 shows the return loss for 8 transmitters, confirming that an adequate bandwidth of about 50 MHz exists in all antennas at around the operating frequency of 2.7 GHz. At the operating frequency of 2.7 GHz, the mean return loss (S11) is -17.9
dB, with a maximum of -14.8 dB and a minimum of -22.1 dB. This plot also confirms that mutual coupling is not causing a significant shift in the resonant frequency of the antennas.

**Polarization and Radiation Pattern**

Figure 8.14 shows the polarization of the local fields surrounding a single bowtie patch antenna. As expected the fields in the broadside direction are linearly polarized and directed along the length of the patch. Fields in the endfire direction are also linearly polarized and are directed perpendicular to the plane of the patch.

Figure 8.15 shows the gain of the antenna for the E-plane ($\phi = 0^\circ$) and the H-plane ($\phi = 90^\circ$), as predicted by HFSS. This pattern is in general agreement with theory, showing strong radiation in the broadside direction, with a null in the endfire direction in the E-plane. Again, it should be noted that the gain values in the positive z direction in Figure 8.15 are attenuated due to the loss in the matching fluid. Therefore, the maximum gain of the antenna is actually somewhat higher than is shown in the plots, but the relative values of the gain as a function of radiation direction (i.e. the shape of the pattern) are accurate.
**Figure 8.13** Measured return loss for 8 transmitters (Blue), centered about the CST predicted resonant frequency of 2.7 GHz. (Red)

**Figure 8.14** Bowtie Patch: Local Field Polarization
Figure 8.15 Far Field Radiation Pattern: (E-plane-Red, H-Plane-Black).

Chamber Transmission Measurements

Figure 8.16 shows a typical S21 plot between antennas on opposing panels, indicating a signal level of about 35 dB, which is much higher than the noise floor of the whole system (about 90 dB).

Finally, the incident signal and the scattered signal (due to a 1 cm ball) for all combinations of antennas are shown in 8.17. The incident signal is the signal measured when there are no objects in the chamber, and it is used for the calibration of the system with the image reconstruction algorithm. The scattered signal is used for reconstruction of the electrical properties of the medium inside the chamber.

Given a measured mean scattered signal of -46 dB and a noise level (from previous
Figure 8.16 Sample S21 plot showing a signal about 35 dB between two opposing panels.

Figure 8.17 Incident signal and scattered signal due to a 1-cm clay ball at the center of the chamber.

characterizations of the system) of less than -90 dB, the SNR of the detected scattered signal from a 1 cm clay ball using a system of this type is approximately 44 dB, which indicates that the antennas are providing a sufficient signal level to perform imaging.
Chapter 9

System Design

The microwave imaging data acquisition system being developed at Duke is composed of a vector network analyzer (which serves as both a microwave source and detector), a switching network of computer-controlled multiplexers and coaxial transmission lines, independent sets of transmitting and receiving antennas, a chamber whose interior contains both a matching medium and the breast to be imaged, and a computer for control and data storage. A schematic of the data acquisition system is shown in Figure 9.1. A detailed description of the design and characterization of the 3D system developed at Duke is presented below.

9.1 Switching Network

A switching system capable of selecting and providing an RF signal path from the Network Analyzer to any combination of 62 Transmit and 63 Receive antennas was needed. The switching system should have the lowest possible insertion loss and provide a good 50 Ω impedance match to maximize signal strength, a high level of isolation between channels to minimize crosstalk, and also provide SMA connectivity
Figure 9.1 Microwave imaging system schematic.

for easy integration into the system. In addition, each switch should be mounted in an enclosure to provide increased noise immunity and structural stability, as well as appropriate power and control connectivity. Finally, earlier work showed that the switches should be absorptive (terminate in 50 Ω load) so as to not act as a secondary radiator.

9.1.1 Multiplexer Design

A surface mount SP4T GaAs absorptive multiplexer chip was selected to meet all of the above criteria (Minicircuits GSWA-4-30DR). Using these chips, two circuit
Figure 9.2 (a) 1x4 Multiplexer (b) 1x16 Multiplexer

Figure 9.3 Transmitter multiplexer switching network schematic.
boards were designed: one capable of switching among 4 outputs (shown in Figure 9.2 (a)) and one capable of switching among 16 outputs (shown in Figure 9.2 (b)). These circuit boards were then connectorized with SMA connectors (for the RF signal lines), DB-15 connectors (for power, ground, and control lines), and finally mounted in individual aluminum enclosures. Using a combination of the simple 1x4 multiplexer and four 1x16 multiplexers, the switching network has the capability to switch a single input among 64 combinations of outputs. Two individual networks (one for the 62 transmitters, the other for the 63 receivers) provided the necessary switching capability for our system. A schematic of the transmitter switching network is shown in Figure 9.3. Control of the switching network is performed through the six control lines $x_0 - x_5$, which are switched using a standard parallel port on a PC. A similar network connects the 63 receivers to the detector port of the network analyzer.

### 9.1.2 Switching Network Characterization

Characterization of the multiplexers included measurements of insertion loss through the multiplexer tree, reflectivity from unselected ports, and isolation performance. The first measurement, insertion loss, was a measure of how much signal is lost through the system (ideally 0dB). The reflectivity measurement was a characteri-
zation of how much signal would be reflected from unselected antennas connected
to the switching system (measurements are taken by selecting 1 antenna pair and
leaving the remaining antennas unselected). For our previous multiplexer, unselected
ports would reflect a very large percentage of the incident signal, possibly affecting
the measurement between the selected antennas, thus, the lower the amount of re-
fectivity, the better. Finally, the isolation characterization is a measure of undesired
coupling between output channels, and should be as low as possible.

Figure 9.4 (a) shows similar insertion loss between the previous and current mul-
tiplexer. While the previous multiplexer shows slightly better insertion loss perfor-
mance, the new multiplexer tree provides an adequately low level of insertion loss for
use in our system. This increase in insertion loss can most likely be attributed to
impedance mismatches associated with our custom PCB design, as well as loss and
mismatch introduced by the in-house fabrication process. Given the improved perfor-
mance in other areas, however, the insertion loss associated with the absorptive mul-
tiplexer tree is unfortunately a necessary cost. Improvements in fabrication methods
and investigation of alternative multiplexer chip technology (e.g. RF MEMS) could
address this issue.
Figure 9.4 (b) shows a comparison of the typical reflectivity of an unselected output port for both the previous and current multiplexer. This plot shows one of the principal advantages of the new absorptive multiplexer chips that were chosen. As can be seen in the plot, the reflectivity was significantly higher for the old multiplexer, in some cases as high as complete reflection. The current multiplexers provide a broadband 50-ohm absorptive match at the unselected ports that reduces the amount of reflection to less than 1 percent at our primary band of operation (around 900MHz). This is critical since it prevents reflections from unused antennas from interfering with the measurement of the active antenna pair.

Finally, the isolation plot in Figure 9.4 (c) shows that the selectivity of our multiplexer switches is around -40dB, which is quite good for this type of multiplexer, but may not be sufficient to ensure that channels carrying strong signals do not obscure the scattering data contained in weaker channels. Design adjustments, such as the inclusion of an additional set of switches to provide an additional layer of isolation (as was done in the Dartmouth switching system) could improve the isolation performance of this system.
Figure 9.4 (a) Insertion Loss (b) Antenna Port Reflectivity (c) Channel Isolation

Future Switching Network

As mentioned above, the switching system that is currently used in the imaging system is made up of transistor-based electronic switches that are inherently lossy and suffer from poor high-frequency performance. An alternative switching technology that is currently under investigation is microelectromechanical system (MEMS) based RF switches. As shown in Figure 9.5, these switches offer vastly improved insertion loss and high-frequency performance over the current switching system. Development of a new switching system that incorporates MEMS technology, along with an additional level of switches to improve channel isolation and enable dual-mode antenna operation, would greatly improve system sensitivity and spatial sampling density.
9.2 Imaging Chambers

9.2.1 Folded Patch Imaging Chamber

Due to the relatively small footprint (1.8cm x 1.8cm) of the folded patch antenna design, it was possible to integrate a uniform 5x5 array of 25 antenna elements onto an 18cm x 18cm printed circuit board panel. The five panels were then assembled into an open top cube to make up the structure of the imaging chamber. Figure 9.6 shows the physical layout of the 62 (dark squares) transmitting antennas and 63 (light squares) receiving antennas. The diagram shows all sides of the five-sided cube unfolded in two dimensions where the center square represents the bottom surface of the cube. Figure 9.7 shows the imaging chamber with the five-sided cube of antenna arrays surrounded by microwave absorbing foam. The microwave absorbing foam
chosen for our experiments was Eccosorb AN-79, which offered reflectivities below -20 dB at 900MHz.

**Bowtie Patch Chamber**

An revised bowtie patch chamber was fabricated (Figure 9.8) that offered some evolutionary improvements to the original bowtie patch chamber described in the previous chapter. First, the updated chamber was fabricated using an improved method that used a CNC milling machine to create four identical and precisely cut antenna panels. The antenna array design was also modified to include a full ground plane backing, which negates the need for absorptive foam and improves the accuracy of the forward modeling of the boundary conditions. Finally, assembly methods were improved by...
Figure 9.7 Imaging chamber with five-sided cube of antenna arrays surrounded by microwave absorbing foam.

Figure 9.8 Revised Bowtie Patch Imaging Chamber (a) Side View (b) Top View
using solder instead of glue to join the five panels that form the chamber, which has eliminated the potential for leaks in the seals between panels. Measurement results in comparison to the HFSS forward model are shown in the modeling section below.

**Clinical Imaging Chamber**

Finally, in order to make it more suitable for clinical use, the bowtie patch imaging chamber was redesigned into a bra-shaped semi-conformal chamber that can be directly attached to the patient’s breast. This clinical imaging chamber is made of a 36-element array of patch antennas fabricated on flexible, low-loss dielectric material (Roger RT/duroid 5870) and integrated with an Alderson Treatment Brassiere cup designed to fit the breast (see Figure 9.10).
9.2.2 Conformal Arrays and Matching Media

As mentioned in the chapter on antenna design, a microwave breast imaging system must be designed to radiate either directly into tissue, or into a matching fluid media designed to have the same electrical properties as mammary tissue. Otherwise, the reflections at the breast interface will be too great, and there will be insufficient signal to reconstruct an image. There are advantages and disadvantages associated with both options. While conformal arrays offer the potential for higher SNR, more compact and cost efficient imaging systems, and a less complex imaging procedure, it is inherently difficult to design antennas that will radiate efficiently into heterogeneous breast tissue with widely varying dielectric properties from patient to patient. Therefore, it is often more practical to use a matching material, despite the additional trouble it may cause.
Figure 9.11 Properties of Acetone (a) Measurement Apparatus (b) $\epsilon_r$ (c) $\sigma$

Acetone Matching Fluid

It was desirable that the matching fluid be a homogeneous, clear, nontoxic, sterile liquid that could be easily pumped in and out of the imaging chamber. To meet these requirements, several fluid choices (mineral oil, glycerine, isopropyl alcohol, and others) were investigated; ultimately, acetone was chosen since it offered the best electrical characteristics (well matched and low loss) as shown in Figure 9.11. However, while acetone is nontoxic and was suitable for use in our investigational experiments in the lab, it is flammable, volatile, has a pungent odor, and is known to be a mild skin irritant, all of which make it impractical for use in a clinical setting.

Custom Dielectric Inserts and Imaging Cart

In order to resolve the problem of patient comfort and safety, an alternative to the acetone matching fluid was needed for use in a clinical setting. Customized low-
loss flexible dielectric inserts of varying sizes that match the dielectric properties
of healthy breast tissue have been designed to address this issue. These dielectric
inserts, made by Emerson and Cuming to a specified dielectric constant between
1 and 30, are inherently low-loss (loss tangent of 0.008) and offer the additional
advantage of comfortably holding the breast steady in a known shape with clearly
defined boundaries (as shown in Figure 9.16 below). Finally, to further improve
patient comfort, an imaging cart has been designed that integrates all of the control
and measurement electronics, RF switching components, and the imaging chamber
into a compact, portable, self-contained clinical imaging system.

This clinical version of the MWI system will have the patient sitting comfortably,
leaning slightly forward to place the breast in the semi-conformal imaging chamber
mounted on the exam cart. The imaging chamber is placed in the slanted side of
the cart (as shown in Figure 9.12) and stays in a fixed position to avoid calibration
errors that can be introduced by movement of the chamber or cables. The coaxial
cables attached to the chamber connect directly to the RF and control electronics
located in the cabinet. The imaging chamber will have replaceable, low-loss flexible
dielectric inserts will come in a variety of sizes and volumes to fit nearly any breast,
Figure 9.12 Microwave Imaging Cart for Clinical Use. Front view shows the central applicator for the imaged breast, and space for either contralateral breast. Side view shows the imaging chamber, head and arm rest, leg gap, and cabinet for control and measurement electronics.

leaving no air gap between the edge of the insert and the skin; they will be easily removable for cleaning after each scan.

Forward Modeling

Currently, both HFSS (finite-element solver) and CST (FDTD solver) are used to model the full imaging chamber. These models can be used to create a numerical characterization of the imaging chamber in the form of polarization data and a Green’s function that will be used to improve the accuracy of the fast forward solvers in the inversion process. Since there are many parameters that are not known exactly
(material properties) it is important to ensure that this characterization is correct.

The simplest empirical method of demonstrating agreement between the simulated and physical chamber is a comparison of the transmission coefficients between a set of antenna pairs.

**Bowtie Patch Chamber Modeling**

A sampling of transmission coefficient comparisons for three antenna pairs located in three different relative positions to one another is shown in Figure 9.13, along with a comparison of a sample return loss measurement for one of the transmitters. In the transmission coefficient comparison plots, a 5dB constant offset has been added to the simulated HFSS results for all cases. This has been done to account for inaccuracies in the current modeling of boundary effects and uncertainty in the dielectric constant of acetone matching fluid.

These issues were resolved in the updated model for the revised bowtie patch antenna chamber (see Figure 9.14, and the constant offset was no longer needed. These results demonstrate that the newly fabricated system is stable and reliable enough to consistently provide measurement results that agree with the predicted values, and that the model is accurate enough for use in supplying the inversion
algorithm with a good description (in the form of Green’s function and polarization data) of how the electromagnetic waves interact with the full environment of the system.

Clinical Bowtie Patch Chamber Modeling

Finally, forward modeling was used to demonstrate the feasibility of imaging breast tissue and detecting early stage breast cancers using the clinical imaging chamber. In order to accomplish this, models of the human breast (both healthy and malignant) were created using published data on the dielectric properties of human tissue and the anatomy of a typical breast (see Figure 9.15). A model of a healthy breast inside the

Figure 9.13 Calibrated HFSS Model: (a) Same Panel Tx/Rx (b) Cross Chamber Tx/Rx (c) Adjacent Panel Tx/Rx (d) Tx Return Loss
clinical imaging chamber is shown in Figure 9.16 and a model containing a 9mm tumor is shown in Figure 9.17. The time averaged electric field results (Figures 9.18, 9.19, 9.20, and 9.21) show that the antennas are radiating efficiently and have achieved sufficient coupling into the breast tissue. In addition, the transmission coefficient results (Figure 9.22) indicate that the scattered field due to small tumors in the breast is also large enough (average value of -56 dB for all combinations of transmit and receive antennas) for potential image reconstruction. Given this scattered field data, the known losses associated with the RF electronics, and the known sensitivity of the measurement device (network analyzer), early detection of small tumors is certainly feasible with the clinical microwave imaging system that has been developed.
Figure 9.15 Human Breast Model With 9mm Tumor

Skin ($\varepsilon_r=3.8$ and $\sigma=1.6$)
Breast Fat ($\varepsilon_r=5.1$ and $\sigma=0.16$)
Glandular Tissue ($\varepsilon_r=20.1$ and $\sigma=0.5$)
9mm Diameter Tumor ($\varepsilon_r=60$ and $\sigma=1$)
Pectoralis Muscle ($\varepsilon_r=52.4$ and $\sigma=1.91$ S/m)
Figure 9.16 Chamber With Healthy Breast Model Containing Skin ($\epsilon_r=3.8$ and $\sigma=1.6$), Subcutaneous Breast Fat ($\epsilon_r=5.1$ and $\sigma=0.16$), Glandular Tissue($\epsilon_r=20.1$ and $\sigma=0.5$), Retromammary Breast Fat(same as Subcutaneous Fat), and Pectoralis Muscle($\epsilon_r=52.4$ and $\sigma=1.91$ S/m).
Figure 9.17 Chamber With Breast Model and 9mm Diameter Tumor with $\varepsilon_r=60$ and $\sigma=1$ S/m (3:1 Dielectric and 2:1 Conductivity Contrast Ratio With Surrounding Glandular Tissue).
Figure 9.18 Time Averaged Electric Field Plots: Empty Chamber (Matching Medium Only).

Figure 9.19 Time Averaged Electric Field Plots: Healthy Breast Model in Chamber.
Figure 9.20 Time Averaged Electric Field Plots: Breast Model With 9mm Diameter Malignant Tumor (3:1 Dielectric and 2:1 Conductivity Contrast Ratio).
Figure 9.21 Time Averaged Electric Field on Surface of Imaging Chamber.
Figure 9.22 Transmission Coefficient as a Function of Receiver Index for Three Sample Transmitters.
Chapter 10

Published Imaging Results

In this chapter, published results from the leading microwave imaging systems are summarized and compared in Figure 10.1. These include the MIST system developed by Hagness et al., the TSAR system developed by Fear and Stuchly, the 2D slice tomographic system developed by Meaney et al., and the 3D inverse scattering systems developed at Duke by Liu et al. Finally, sample images from the Duke 3D MWI systems are shown.

10.1 3D Tomographic Images from the Duke MWI Systems

Figures 10.2 and 10.3 show sample imaging results from the Duke 3D scanning antenna system, and Figure 10.4 shows a sample imaging result from an earlier Duke 3D fixed array prototype system that was designed to operate in air. These results demonstrate that detection and imaging of inclusions with dielectric contrast is possible using the 3D systems and methods developed at Duke.
<table>
<thead>
<tr>
<th>PI</th>
<th>Location</th>
<th>Antenna</th>
<th>Elements</th>
<th>Imaging Method</th>
<th>Images</th>
<th>Detection/Resolution</th>
</tr>
</thead>
<tbody>
<tr>
<td>Liu</td>
<td>Duke</td>
<td>Patch (CW)</td>
<td>32-125 (&lt;2 Min Scan)</td>
<td>3D Inversion Methods</td>
<td>Multiple Phantoms in Layered Media (Skin, Fluid, Tumor)</td>
<td>Multiple Phantoms &lt;5 mm Diameter (Homogenous)/&lt;1 cm (Layered), &lt;2 cm Separation</td>
</tr>
<tr>
<td>Meaney</td>
<td>Dartmouth</td>
<td>Monopole (Multifreq CW)</td>
<td>16 (15 Min Scan)</td>
<td>2D Inversion Methods</td>
<td>In Vivo Imaging 7 Slices (2D) at 7 Frequencies</td>
<td>130 Subjects &gt;1 cm Detection</td>
</tr>
<tr>
<td>Hagness</td>
<td>Wisconsin</td>
<td>UWB Pyramidal Horn (Pulsed)</td>
<td>Single Element Scanned 49 Positions</td>
<td>Space-Time Beam-forming (MIST) combined with Statistical Methods (GLRT)</td>
<td>Multiple Phantoms in Layered Media (Skin, Oil, Tumor)</td>
<td>Multiple Phantoms 4 mm in Diameter 2 cm Separation</td>
</tr>
<tr>
<td>Fear</td>
<td>Calgary</td>
<td>UWB Wu-King Monopole (Pulsed)</td>
<td>Single Element Scanned 49 Positions</td>
<td>Quasi-3D, Tissue Sensing Adaptive Radar (TSAR) and MIST</td>
<td>Phantom Embedded in Layered Media (Skin, Oil, Tumor)</td>
<td>Single Phantom 1 cm in Diameter (Layered Media) 4 mm in Diameter (Homogenous)</td>
</tr>
<tr>
<td>Klemm</td>
<td>Bristol</td>
<td>UWB Stacked Conformal Patch (Pulsed)</td>
<td>16 Elements in 4x4 Array</td>
<td>Delay and Sum Beam-forming</td>
<td>Phantoms in Homogenous Liquid</td>
<td>Single Phantom &lt;6 mm Diameter</td>
</tr>
<tr>
<td>Pastorino</td>
<td>Genoa</td>
<td>No Published Results Found</td>
<td>No Published Results Found</td>
<td>Genetic Algorithm</td>
<td>Only Simulated Imaging Results Found</td>
<td>Sub-wavelength Detection (Simulated)</td>
</tr>
</tbody>
</table>

**Figure 10.1** Comparison of Published Imaging Results.

**Figure 10.2** 5 mm Dielectric Phantom Imaged with 3D Scanning System. (a)DTA-BIM (b)BCGS-BIM
Figure 10.3 Multiple 5 mm Dielectric Phantoms Resolved with 3D Scanning System. (a),(c): DTA-BIM (b),(d): BCGS-BIM

Figure 10.4 Single Dielectric Phantom Imaged with 3D Fixed Array System (Air Background).
Chapter 11

Remaining Work and Conclusions

Examining the results of the four leading active microwave imaging research groups, the future of MWI as a potential alternative detection modality for early stage breast cancer appears strong. The successful detection and imaging of tumors demonstrated in the Dartmouth clinical trials is encouraging, especially when combined with all of the potential improvements offered by a full 3D scattering approach. The successful inversion of phantoms from experimental data in the 3D imaging systems at Duke have demonstrated that these potential improvements are real. What remains is to finish implementing the improvements in the hardware system (specifically the switching system) and constructing the Green’s function from the completed forward model of the 3D prototype clinical system. From there, optimizations based on phantom imaging experiments can be performed, leading to clinical trials of the Duke 3D microwave breast imaging system.

Finally, it is important to remember that in addition to microwave imaging, several other alternative breast cancer detection modalities are actively being pursued,
including optical imaging methods, EIT, PET/SPECT systems, radiometric imaging systems (also sometimes called passive microwave imaging since antennas are being used to measure increased thermal energy emanating from a tumor location), and systems based on biological markers (BRCA1 and BRCA2), as well as many hybrid methods of those already mentioned. Each of these methods offers its own advantages and disadvantages, and certain methods will emerge as superior to mammography in specific ways, likely leading to the use of multiple imaging modalities for optimal detection. Ultimately, all of these efforts should give hope that something close to the ideal breast cancer detection system described by IOM may soon be within reach.
Appendix A

Definition of S-Parameters

In microwave imaging, it is useful to characterize the electromagnetic signals fed to the transmitting antennas and captured by the receiving antennas by scattering parameters (or simply S-parameters). The scattering parameters are of the form $S_{nm}$ (which is the ratio of received signal as measured at port $n$ to the incident signal from port $m$). $S_{nn}$ is commonly known as the reflection coefficient at port $n$ and $S_{nm}$ as the transmission coefficient from port $m$ to port $n$.

Considering just two antennas, one transmitting and one receiving, the scattering parameters may be simply expressed. The electric field intensity at the transmitter location ($E_1$) of the imaging chamber and the electric field intensity at the receiver location ($E_2$) are made up of incident and reflected (scattered) components.

This relationship is expressed as,

\[ E_1 = E_{1i} + E_{1r} \quad \text{(A.1)} \]

and

\[ E_2 = E_{2i} + E_{2r} \quad \text{(A.2)} \]
where $E_{1i}$ is a wave incident from antenna 1 into the chamber, $E_{1r}$ is reflected into antenna 1 from the chamber, $E_{2i}$ is incident from antenna 2 into the chamber and $E_{2r}$ passes from the chamber and into antenna 2.

Considering the polarization vectors $p_1$ and $p_2$ of antennas 1 and 2, the signal received at the two antennas can be written as $E_1 = E_1 \cdot p_1$ and $E_2 = E_2 \cdot p_2$, respectively.

Since the incident and scattered fields completely determine the transmitted and received signals, it is convenient to express the scattered fields as functions of the incident fields as,

$$E_{1r} = S_{11}E_{1i} + S_{12}E_{2i} \quad (A.3)$$

and

$$E_{2r} = S_{21}E_{1i} + S_{22}E_{2i} \quad (A.4)$$

where the scattering parameters are defined as:

$$S_{11} = E_{1r}/E_{1i}|_{E_{2i}=0}, \quad S_{12} = E_{1r}/E_{2i}|_{E_{1i}=0}, \quad S_{21} = E_{2r}/E_{1i}|_{E_{2i}=0}, \quad S_{22} = E_{2r}/E_{2i}|_{E_{1i}=0} \quad (A.5)$$

Thus, $S_{11}$ is the reflection coefficient at antenna 1, under the condition that antenna 2 is terminated in the impedance of its connecting cable (usually 50 $\Omega$), so that no
signal enters the region from antenna 2. Under the same condition at antenna 2, $S_{21}$

is the forward transmission coefficient of signals from antenna 1 to antenna 2.

Likewise, $S_{22}$ is the reflection coefficient at antenna 2, under the condition that antenna 1 is terminated in the impedance of its connecting cable. Under the same condition at antenna 1, $S_{12}$ is the reverse transmission coefficient of signals from antenna 2 to antenna 1.
Appendix B

Definition of Antenna Terms

The following are some important antenna terms used throughout the paper, as defined by the IEEE, and found in [51].

Antenna Aperture

- a surface on or near the antenna on which assumptions about the fields may be made for calculating the field at external points.

Antenna Efficiency

- the ratio of the power radiated, $P_r$, to the power fed to the antenna $P_T$:

$$\eta = \frac{P_r}{P_T} \times 100 \quad (B.1)$$

Bandwidth

- the frequency range over which the antenna characteristics conform to a desired standard. For microstrip antennas, the bandwidth is determined by a permissible value of SWR, typically 1:2.
Directivity

- the ratio of the maximum radiation intensity to the average radiation intensity.

E-Plane and H-Plane Pattern

- the E-plane Pattern is the radiation pattern in the plane containing the electric field vector and the direction of maximum radiation. This is a plot of \( R(\theta) \) for \( \phi = 0 \). The H-plane Pattern is the radiation pattern in the plane containing the magnetic field vector and the direction of maximum radiation. This is a plot of \( R(\theta) \) for \( \phi = 90^\circ \).

Gain

- the gain of antenna is the product of its directivity and efficiency.

Input Impedance

- the impedance presented by an antenna at its terminals

\[
Z_{in} = \frac{V^2}{2P_T} \quad (B.2)
\]
Polarization of an Antenna

- the time varying direction of the electric field vector of the wave radiated by the antenna in a particular direction. When the direction is not specified, it is taken to be the direction of maximum radiation.

Radiation Pattern

- the directional dependence of radiation from an antenna.

Resonant Frequency

- a frequency at which the input impedance of an antenna has no reactive component.
Bibliography


Biography

John Patrick Stang was born in Kinston, North Carolina on May 10, 1982 to parents Richard and Sharon. He grew up in Greenville, North Carolina, attending St. Peter’s School from 1987-1996 and J. H. Rose High School from 1996-2000, where he met his future wife, Taylor. John graduated from Duke University with a Bachelor of Science in electrical engineering in 2004 and went on to complete a Master of Science in electrical engineering at Duke in 2007. His publications include the conference papers “A Tapered Microstrip Patch Antenna Array for Microwave Breast Imaging” and “A Compact CPW-Fed Microstrip Patch Antenna Array for Unmanned Aerial Vehicle RADAR”, where he was first author, as well as two additional articles and a book chapter in which he was a contributing author. He received the Charles R. Vail award from Duke University in 2007. Following the completion of his Ph.D., John will be relocating to New York City.