Ultra-wideband (UWB) microwave imaging for breast cancer detection : UWB experiments on breast phantoms

Lai, Joshua Chong Yue

2009

Lai, J. C. Y. (2009). Ultra-wideband (UWB) microwave imaging for breast cancer detection : UWB experiments on breast phantoms. Master's thesis, Nanyang Technological University, Singapore.

<https://hdl.handle.net/10356/42107>

<https://doi.org/10.32657/10356/42107>

Ultra-Wideband (UWB) Microwave Imaging for Breast Cancer Detection: UWB Experiments on Breast Phantoms

Joshua Lai Chong Yue

School of Electrical & Electronic Engineering

A Thesis submitted to the Nanyang Technological University
in fulfillment of the requirement for the degree of
Master of Engineering

2009

Acknowledgements

I would like to express my utmost gratitude to my supervisor, Assoc. Prof. Soh Cheong Boon, for his guidance and encouragement during the course of the thesis. I am very grateful for the opportunities that he has given me to learn and explore, and truly appreciate the time and effort that he spent to guide me during the research.

I also want to thank Assoc. Prof. Low Kay Soon, Assoc. Prof Erry Gunawan from the School of Electrical and Electronics Engineering. They have brought many insights and new ideas to my research project. My discussions with them have been very helpful.

I owe thanks to Steven Poh and Chua Lye-Whatt from Thales research group for borrowing the UWB antennas for the experiments. Without the antennas from them, the experiments are almost impossible to be conducted.

At last, I would like to thank Dr. Chen YiFan, Lin Lin Thi, Arash Maskooki, Teo JianQi, Loh Yu Han, Cher Zhong Ming, Chua Seng Boon, and Teh Jin Pin in Biomedical Engineering Research Center for making the lab cheerful and lively.

Table of Contents

Acknowledgement.....	i
Table of Contents.....	ii
Abstract	iv
List of Figures	v
List of Tables	viii
Chapter 1: Introduction.....	1
1.1 Background.....	1
1.1.1 <i>Ultra-wideband</i>	1
1.1.2 <i>Breast Cancer Statistics</i>	2
1.1.3 <i>Existing Imaging Modalities for Breast Cancer Detection</i>	3
1.2 Motivations	5
1.3 Objectives	5
1.4 Major Contributions of the Thesis	6
1.5 Organization of the Thesis.....	7
Chapter 2: Literature Review	8
2.1 Dielectric Properties of Breast Tissues	8
2.2 Microwave Breast Imaging.....	12
2.3 Algorithms for UWB based Breast Cancer Detection	14
2.3.1 <i>Image Formation</i>	14
2.3.2 <i>Early Time Artifact Removal</i>	14
2.4 UWB Experiments for Breast Cancer Detection	15
2.5 Breast Phantom Materials.....	19
Chapter 3: Methodology	20
3.1 Experimental setup	20
3.1.1 <i>Equipment used in Experiment</i>	21
3.1.2 <i>Planar Rectangular and Circular Antenna Arrays</i>	25
3.2 Signal Processing.....	26
3.3 Calibration experiments	32
Chapter 4:	
Experimental Issues and Compensation Methods	35
4.1 Pulse Instability Artifacts and Compensation.....	36
4.2 Loss Compensation.....	41
4.2.1 <i>Radial Spreading Loss</i>	42
4.2.2 <i>Antenna Pattern Loss</i>	42

4.2.3 Medium Dielectric Loss	44
4.2.4 Effects of Loss Compensation	45
4.3 Filtering and Correlation.....	46
4.3.1: Filtering.....	46
4.3.2: Correlation	48
4.4 Ring Artifact	52
4.4.1 Experiment on Ring Artifact	54
4.5 Dynamic Range of Real-Time Oscilloscope	56
4.5.1 Vertical Resolution of Real-Time Oscilloscope	57
4.5.2 Dynamic Range of Real-Time Oscilloscope	58
4.6 Signal and Image SNR.....	61
4.6.1 SNR Improvement with Larger Averaging and Antenna Number	62

Chapter 5:

Fabrication and Dielectric Measurement of Breast Phantoms 69

5.1 Probe Performance.....	69
5.2 Time Effect on Measurement Accuracy	72
5.3 Phantom Materials in Cylindrical Container	74
5.4 Phantom Materials in Flat Rectangular Container.....	78
5.5 Fabrication of Breast Phantoms.....	81
5.6 Dielectric Properties of Homogeneous Breast Phantoms	83
5.7 Dielectric Properties of Heterogeneous Breast Phantoms	88
5.8 Analysis of Dielectric Variability	92

Chapter 6:

UWB Imaging on Realistic Breast Phantoms 99

6.1 Experiment Design	99
6.2 Imaging of Homogeneous Breast Phantoms.....	104
6.3 Imaging of Heterogeneous Breast Phantoms.....	108
6.4 Limiting Factors for Tumor Detectability	116
6.5 Tumor Response Analysis	118
6.6 Estimation of Detection Resolution.....	121

Chapter 7: Conclusion and Future Work 125

References..... 127

Appendices..... 131

Abstract

The potential of UWB breast cancer detection has been demonstrated with simple experiments on homogenous breast phantoms which do not represent the tissue dielectric properties of heterogeneous human breast. This thesis attempts to bridge the gap by conducting experiments on pulse-based UWB microwave breast imaging with realistic tissue mimicking breast phantoms. Experimental issues which are important for the development of pulse-based UWB breast cancer detection prototype and the compensation methods are studied. These includes pulse instability, loss compensation, narrowband frequencies interference, limited dynamic range, phantom positioning error, averaging number and antenna number. Detection capability is shown to be significantly improved by the compensation methods.

The breast phantoms used in the experiments are realistic and comparable to human breast in term of dielectric properties. Unlike previously reported experiments that use lower (compared to human breast tissue) dielectric homogeneous materials as the breast medium, the experiments are significantly different in two ways. Firstly, dielectric loss and dielectric constant of the breast phantoms used in our experiments are close to actual human breasts. Secondly, heterogeneous breast phantoms are fabricated by mixing phantom materials of different dielectric constants to simulate the glandular tissue clutters in adipose tissue matrix of the human breast.

More realistic experiments are designed based on the new results from large scale dielectric measurements on normal and malignant breast tissues. Extensive experiments are conducted to simulate the wide variability in dielectric properties of human breast. Experiment results show that the experimental setup is capable of detecting 2 mm tumor in breast phantoms.

List of Figures

- Figure 1.1: FCC frequency mask for UWB system
- Figure 2.1: Dielectric constant and conductivity for normal and malignant breast tissues
- Figure 2.2: Methods of microwave breast imaging
- Figure 2.3: Experimental setup for breast cancer detection at University of Winconsin
- Figure 2.4: Experimental setup for breast cancer detection at University of Calgary
- Figure 3.1: Overall Experimental Setup
- Figure 3.2: Picosecond 3500D Impulse Generator
- Figure 3.3: Agilent DSO81204B Real-Time Oscilloscope
- Figure 3.4: Agilent N5230A Vector Network Analyzer
- Figure 3.5: High precision x-y-rotary experiment stage
- Figure 3.6: Front view and side view of Thales UWB antenna
- Figure 3.7: Radiation pattern of Thales UWB antenna
- Figure 3.8: S-parameters of Thales UWB antenna
- Figure 3.9: Experiment setup for circular antenna array
- Figure 3.10: Flow chart of Delay-and-Sum algorithm
- Figure 3.11: Signals at each step of Delay-and-Sum algorithm
- Figure 3.12: Energy calculation and image plotting
- Figure 3.13: Simple breast phantom of oil (Left) and screws used as the tumor (Right)
- Figure 3.14: Imaging results of calibration experiments with tumor simulants
- Figure 4.1: Signals before and after pulse instability compensation
- Figure 4.2: Zoom in view of signals given in Figure 4.1
- Figure 4.3: Resultant noise after averaging method for signals with and without pulse instability compensation
- Figure 4.4: The effects of pulse instability compensation on breast phantom images
- Figure 4.5: Simulated received pulse amplitude considering only radial spreading loss
- Figure 4.6: Normalized gain of Thales antenna with half-power beamwidth of 60 degrees
- Figure 4.7: Antenna gain in the scanning region for transmit and receive antennas
- Figure 4.8: Simulated received pulse amplitude considering antenna pattern loss
- Figure 4.9: Image with (left) and without (right) radial spreading compensation
- Figure 4.10: Signals and power spectrums before and after filtering
- Figure 4.11: Images formed without filtering (left) and with filtering applied (right)

Figure 4.12: Tumor response template and its power spectrum

Figure 4.13: Comparison for images without correlation (left) and with correlation (right)

Figure 4.14: Ten filtered images without correlation (top) and corresponding filtered images with correlation (bottom)

Figure 4.15: SNR versus number of antennas

Figure 4.16: Images formed with tumor located 1 cm to 6 cm from center

Figure 4.17: Images of breast phantom 1 mm off-center (left) and breast phantom less than 0.5 mm off-center (right)

Figure 4.18: Resultant noise after averaging method for best positioning and 1mm displacement from center

Figure 4.19: Images for phantom with best positioning (top-left), 1mm displacement (top-right), 2mm displacement (bottom-left), and 3mm displacement (bottom-right)

Figure 4.20: Histogram of a pulse recorded from oscilloscope

Figure 4.21: Ideal tumor response constructed with 2-bit, 4-bit, and 6-bit of resolutions

Figure 4.22: Constructed pulse to test dynamic range of 2^{11} -bits

Figure 4.23: RMS noise versus averaging number for different pulse attenuation settings

Figure 4.24: Image SNR versus Signal SNR for different antenna numbers

Figure 4.25: Images obtained with ten sets of noise processed with antenna number of 360 and RMS noise level of 2.2 mV

Figure 4.26: Images obtained with the same set of noise processed with antenna number of 360 and RMS noise level from 2.2 mV to 0.10 mV

Figure 4.27: Images obtained with the same set of noise processed with antenna number of 12, 24, 45, 90, 180, 360 and RMS noise level of 0.45 mV

Figure 5.1: Open-ended coaxial probe used for dielectric properties measurement

Figure 5.2: Dielectric constant and conductivity of water

Figure 5.3: Dielectric constant and conductivity of methanol, ethanol, and propanol

Figure 5.4: Water with and without electronic calibration auto-refresh function on

Figure 5.5: Phantom material dielectric constant at 5 GHz measured within 12 hours

Figure 5.6: Description of dielectric measurement procedure

Figure 5.7: Dielectric constants for phantom materials with different concentration of oil

Figure 5.8: Dielectric sheet in a flat rectangular container

Figure 5.9: Dielectric constants for four phantom material sheets with 50% oil

Figure 5.10: Low dielectric matrix material and high dielectric clutter material

- Figure 5.11: Box plot and histogram for breast phantom Homo-50a
- Figure 5.12: Box plot and histogram for breast phantom Homo-50b
- Figure 5.13: Box plot and histogram for breast phantom Homo-65a
- Figure 5.14: Box plot and histogram for breast phantom Homo-65b
- Figure 5.15: Box plot and histogram for breast phantom Homo-80a
- Figure 5.16: Box plot and histogram for breast phantom Homo-80b
- Figure 5.17: Box plot and histogram for breast phantom Hetero-17
- Figure 5.18: Box plot and histogram for breast phantom Hetero-25
- Figure 5.19: Box plot and histogram for breast phantom Hetero-33
- Figure 5.20: Box plot and histogram for breast phantom Hetero-50
- Figure 5.21: Box plot and histogram for breast phantom Hetero-60
- Figure 5.22: Box plot and histogram for breast phantom Hetero-65
- Figure 5.23: Box plot and histogram for breast phantom Hetero-70
- Figure 5.24: Mean and variance for each group of measurements
- Figure 5.25: Box plot for each surface of measurement for a homogeneous phantom
- Figure 6.1: Dielectric constant of breast tissues
- Figure 6.2: Variability of breast tissues dielectric constant
- Figure 6.3: Materials used as tumor simulant in breast phantom
- Figure 6.4: Imaging results for breast phantom Homo-50
- Figure 6.5: Imaging results for breast phantom Homo-65
- Figure 6.6: Imaging results for breast phantom Homo-80
- Figure 6.7: Imaging results for breast phantom Hetero-17
- Figure 6.8: Imaging results for breast phantom Hetero-25
- Figure 6.9: Imaging results for breast phantom Hetero-33
- Figure 6.10: Imaging results for breast phantom Hetero-50
- Figure 6.11: Imaging results for breast phantom Hetero-60
- Figure 6.12: Imaging results for breast phantom Hetero-65
- Figure 6.13: Imaging results for breast phantom Hetero-70
- Figure 6.14: Pixel intensity versus dielectric contrast for homogeneous breast phantoms
- Figure 6.15: Pixel intensity versus percentage of clutters for heterogeneous breast phantoms
- Figure 6.16: Pixel intensity versus dielectric variability for heterogeneous breast phantoms

List of Tables

- Table 1.1: Year 2008 women cancer statistics in the United States
- Table 1.2: Survival rate of breast cancer patients diagnosed at different stages
- Table 1.3: Typical numbers of women with mammography screening
- Table 2.1: Dielectric constants at 5 GHz of normal breast tissues from Lazebnik's large scale studies
- Table 2.2: Dielectric constants at 5 GHz of cancer and normal breast tissues from Lazebnik's large scale studies and previously assumed values
- Table 2.3: Dielectric constant and conductivity of breast phantom materials at 6 GHz used by University of Wisconsin as compared to actual human tissues
- Table 2.4: Dielectric constant and conductivity of breast phantom materials at 4 GHz used by University of Calgary as compared to actual human tissues
- Table 4.1: Simulated and experimental noise amplitude after pulse instability compensation at different sampling rates
- Table 4.2: Alpha and Attenuation for different mediums at 5 GHz
- Table 4.3: SNR and detectability for different numbers of antennas per phantom rotation
- Table 4.4: Averaged RMS and peak-to-peak amplitude of resultant noise after averaging method for different positioning error
- Table 4.5: Bits of resolution for different averaging settings and different vertical scales
- Table 4.6: Expected and recorded peak of incident pulse for dynamic range
- Table 4.7: RMS and peak-to-peak amplitude of noise for different averaging number with maximum pulse amplitude 8V (attenuation setting 0 dB)
- Table 4.8: Image SNR for different averaging number and antenna number. Signal SNR is calculated with tumor response RMS of 0.2567 mV
- Table 5.1: Water dielectric constant at 0.5 GHz for three coaxial probes
- Table 5.2: Debye parameters for standard materials
- Table 5.3: Dielectric constant for three standard materials at 5 GHz
- Table 5.4: Homogeneous breast phantoms fabricated with different percentage of oil
- Table 5.5: Compositions of seven heterogeneous breast phantoms fabricated
- Table 5.6: Variance of different groups for phantom materials
- Table 5.7: Variance of different groups for sheets with 50% oil
- Table 5.8: Mean and deviation from grand mean (16.36) for sheet 1 to sheet 4

Table 5.9: Variance for homogeneous breast phantoms

Table 5.10: Variance for heterogeneous breast phantoms

Table 5.11: Variability for homogeneous breast phantoms

Table 5.12: Variability for heterogeneous breast phantoms

Table 6.1: Dielectric constants at 5 GHz and variability for fabricated homogenous and heterogeneous breast phantoms

Table 6.2: Dimension and dielectric constant of materials used as tumor simulant

Table 6.3: Detectability of tumor in different homogeneous phantoms

Table 6.4: Detectability of tumor in different heterogeneous phantoms

Table 6.5: Artifact amplitudes due to system limitation

Table 6.6: Averaging artifacts due to algorithm limitation

Table 6.7: Clutters responses due to breast phantom limitation

Table 6.8: Ratios of summed signal amplitude over max signal amplitude for different tumor simulants

Table 6.9: SNR for tumors of different sizes due to system limitation

Table 6.10: SNR for tumors of different sizes due to algorithm limitation

Table 6.11: Breast limited SCR for clutters of different dielectric constants

Table 6.12: Breast limited SCR for different percentage of clutters

Chapter 1: Introduction

Breast cancer is the most common cancer among women. Mammography which is the main detection method has known drawbacks such as ionizing radiation, painful breast compression and high false positive and false negative rates. Magnetic resonance imaging (MRI) and ultrasound are in various stages of use in diagnosis but each of these imaging modalities has its limitations. MRI exhibits highly variable specificity depending on the interpretation technique used and the patient population. Moreover, the cost of MRI scanners is high and the availability of the technique and the expertise required for interpretation is very limited. Ultrasound is highly non-specific for breast lesions smaller than 1 cm in size. Early stage breast cancer detection and timely medical intervention is critical to patients' survival. Therefore there is a clinical need for a new complimentary low cost medical imaging method to improve upon the detection of early stage breast cancer.

In recent years, there is much research interest in using microwave imaging technique for breast cancer detection. The physical basis is the significant contrast in the dielectric properties of normal and malignant breast tissues.

1.1 Background

1.1.1 Ultra-wideband

Ultra-wideband (UWB) uses the microwave spectrum ranging from 3.1 GHz to 10.6 GHz as shown in Figure 1.1 allocated by the United States Federal Communications Commission (FCC) in 2002 [1] where the bandwidth is larger than 500 MHz or 20% of the center frequency. In time domain, UWB is a very short electromagnetic pulse of sub nanosecond duration.

The new FCC regulation has motivated the research in UWB for next generation communication technology such as Personal Area Network and wireless USB. UWB is also used in radar for subsurface sensing applications such as mines detection, locating utility lines, examining archaeology sites, and studying ground water [2]. UWB radar has also attracted a lot of interest for early stage breast cancer detection [11]–[30].

Figure 1.1: FCC frequency mask for UWB system [1]

1.1.2 Breast Cancer Statistics

Breast cancer is the most common cancer and the second leading cause of cancer death in women as shown in Table 1.1. In the United States, 182460 new breast cancer cases are expected to be found with 40480 women expected to die from breast cancer in 2008. These numbers represent 26% of estimated new cancer cases and 15% of estimated deaths for all types of cancer in women [3].

Table 1.1: Year 2008 women cancer statistics in the United States

Estimated new cases		Estimated deaths	
breast cancer	26%	lung cancer	26%
lung cancer	14%	breast cancer	15%
colorectal cancer	10%	colorectal cancer	9%

The United States National Cancer Institute estimates that 12.7% of women born today will be diagnosed with breast cancer at some time in their lives [4]. In other word, one out of eight women will develop breast cancer during their life time. Early detection of breast cancer is proven to increase survival rate of breast cancer patients [5, 6] as shown in Table 1.2.

Table 1.2: Survival rate of breast cancer patients diagnosed at different stages

Breast Cancer Stage	0	I	IIA	IIB	IIIA	IIIB	IV
5-year Survival Rate	100%	100%	92%	81%	67%	54%	20%

1.1.3 Existing Imaging Modalities for Breast Cancer Detection

Mammography

Mammography is the current gold standard for breast cancer detection. Mammography produces high resolution image of x-rays attenuation by breast tissues. However, the contrast between malignant and normal breast tissues at x-ray frequencies is low [7].

Although mammography has been widely accepted for many years, its limitations are well known [10]. As shown in Table 1.3, on average mammography finds 70 women with suspected tumor among 1000 women who have mammography screening, but only 3.5 women have breast cancer. Mammography suffers from high false-positive [8] and high false-negative rates [9], and its detection capability depends on radiologists' interpretation. Other drawbacks of mammography are the use of ionizing radiation on the human body and requiring breast compression during the scanning procedure.

Table 1.3: Typical numbers of women with mammography screening

Number of women	Medical procedure
1000	Screening mammography (930 found nothing)
70	Diagnostic mammography (60 found benign)
10	Biopsy (6.5 found benign)
3.5	Cancer Confirmed

Ultrasound Imaging

Ultrasound is usually used in directing image guided breast cancer biopsy because it can operate in continuous mode and it poses no radiation hazard. It is also used to assist breast cancer diagnosis after suspicious mammogram result where it can help in differentiating cysts (typically benign) from solid masses (often cancerous).

Ultrasound suffers from high attenuation and ultrasound images have much lower resolution as compared to mammography [7]. Reflection coefficient of ultrasound at the fat and muscle boundary is very small resulting in the small contrast between normal and malignant breast tissues.

Magnetic Resonance Imaging (MRI)

MRI is excellent at differentiating water content in the body. It does not suffer from attenuation problem because of the different physical principle used to generate images.

Unlike mammography which uses only x-ray attenuation to create images, MRI has many properties that can be used to enhance the image contrast. By varying the scanning parameters, tissue contrast can be enhanced to detect different tissues. However, MRI is too expensive to be used for breast cancer screening.

1.2 Motivations

Breast cancer is a significant health issue for women and affects one in every eight women [4]. Mammography which is the current gold standard has known drawbacks such as ionizing radiation, painful breast compression, high false positive and false negative rates. Thus, new approaches for breast cancer screening and medical imaging are needed.

The research on UWB for breast cancer detection is motivated by the high dielectric contrast between normal and malignant breast tissues at UWB frequencies. The large difference in dielectric properties causes the malignant tumor to become a strong microwave scatterer which can be detected. The method is attractive to patients because both ionizing radiation and breast compression are avoided, potentially leading to safer and more comfortable exams.

In addition, UWB signal attenuation in breast tissues is low [11] which enables the detection of tumors in all types of breasts. The UWB signal with much lower power than cell phone radiation makes it safe to be used on human. Therefore, UWB breast cancer detection has the potential to be a low cost complementary imaging technique to existing imaging modalities to improve the detection of early stage breast cancer.

1.3 Objectives

The thesis presents experimental investigation of UWB microwave imaging for breast cancer detection using tissue mimicking breast phantoms. The primary goals of this research include:

- To set up the experimental stage for scanning the breast phantom using UWB signals and collecting the backscattered signals
- To design and fabricate tissue mimicking breast phantoms with dielectric properties similar to human breasts
- To conduct UWB breast cancer detection imaging experiments on heterogeneous breast phantoms

1.4 Major Contributions of the Thesis

UWB imaging experiments which have demonstrated the potential for the detection of early stage breast cancers were conducted using homogenous breast phantoms. The clutters from the heterogeneous nature of human breast were ignored. The UWB imaging experiments in the thesis uses more realistic heterogeneous breast phantoms to investigate the signal-to-clutter ratio which is important for detection of breast cancer.

The breast phantoms used in the experiments are realistic and comparable to human breast in term of dielectric properties. Unlike previously reported experiments that use lower dielectric homogeneous materials as the breast medium, the experiments are significantly different with dielectric properties and variability of the breast phantoms close to actual human breasts.

Thus far, UWB imaging experiments were based on frequency domain approach using a vector network analyzer. A pseudo UWB pulse was synthesized from continuous waves with frequencies in the UWB band using the inverse fourier transform. However, wireless USB uses the sub nanosecond UWB pulses and low cost UWB pulse transceivers are expected to be available for imaging applications. Therefore, the thesis conducts UWB pulse-based imaging experiments to investigate the feasibility of pulse-based UWB to detect early stage breast cancer.

Experimental issues and hardware limitations which are important for the development of pulse-based UWB breast cancer detection prototype and the compensation methods are studied. These includes pulse instability, loss compensation, narrowband frequencies interference, limited dynamic range, phantom positioning error, averaging number and antenna number. Detection capability is shown to be significantly improved by incorporating the compensation methods.

1.5 Organization of the Thesis

Chapter 1 is the introduction. UWB, breast cancer statistics, and existing imaging modalities for breast cancer detection are reviewed. Motivations, objectives and major contribution of the thesis are highlighted.

Chapter 2 is the literature review. Breast tissue dielectric properties, microwave imaging techniques and algorithms used in breast cancer detection are reviewed. UWB experiments done by other research groups and their limitations are highlighted. The materials used for designing breast phantom are discussed.

Chapter 3 discusses the UWB experimental setup and the equipment used in the experiments. Calibration experiments to test the detectability of tumor are discussed.

Chapter 4 discusses the experimental issues encountered and the compensation methods implemented. These include pulse instability, loss compensation, narrowband frequencies interferences, limited dynamic range of real-time oscilloscope, phantom positioning error, number of averaging and antennas used.

Chapter 5 discusses the dielectric properties measurements of tissue mimicking phantom materials. Fabrication and dielectric measurements of homogeneous and heterogeneous breast phantoms are described.

Chapter 6 discusses UWB imaging experiments with homogeneous and heterogeneous breast phantoms. The UWB imaging detection resolution is investigated.

Chapter 7 presents the conclusion and future work.

Chapter 2: Literature Review

2.1 Dielectric Properties of Breast Tissues

Small scale studies have been conducted to measure the dielectric properties of normal and malignant breast tissues. Chaudhary et al. [36] measured the dielectric properties of excised normal and malignant breast tissue specimens from 3 MHz to 3 GHz. The dielectric properties of infiltrating breast carcinoma and selected surrounding nonmalignant tissue in the 20 kHz - 100 MHz range are measured by Surowiec et al. [37]. Joines et al. [38] measured the dielectric properties of freshly excised tissues from several organs, including breasts, over the frequency range 50 MHz - 900 MHz. These small scale studies have shown that the dielectric constant and conductivity for malignant breast tissue is three or more times greater than that of normal breast tissue.

However, all the small scale studies are conducted at frequencies below 3 GHz, whereas UWB frequencies lie between 3.1 GHz to 10.6 GHz. In most theoretical studies on UWB for breast cancer detection [11, 15, 20, 26], the dielectric properties of breast tissues are extrapolated beyond 3 GHz using the Debye equation based on the previous low frequencies studies as shown in Figure 2.1. Dielectric constant of breast tissues at UWB frequencies is commonly assumed to be $9 \pm 10\%$.

Figure 2.1: Dielectric constant and conductivity for normal and malignant breast tissues at microwave frequencies. Shaded area is UWB. Figure taken from [16]

Recently, Lazebnik et al. [48, 49] conducted two large scale studies involving close to three hundred patients with breast tissues samples from reduction surgeries and cancer surgeries. Breast tissues are separated into three groups for statistical analysis, where group 1 contained all samples with 0–30% adipose tissue (the high-water-content group), group 2 contained all samples with 31–84% adipose tissue, and group 3 contained all samples with 85–100% adipose tissue (the low-water-content group). Dielectric properties for the three groups are summarized by Cole-Cole parameters for median, 25th percentile, and 75th percentile. The results are summarized in Table 2.1, where the dielectric constants at 5 GHz are shown. The standard deviations are calculated from the interquartile range by assuming normal distribution of the samples.

$$\text{Interquartile Range (IQR)} = 1.35 \times \text{Standard Deviation (SD)}$$

Table 2.1: Dielectric constants at 5 GHz of normal breast tissues from Lazebnik's [48,49] large scale studies

	Median	IQR	SD
From reduction surgeries			
Group 1	44.2	14.8	11.0
Group 2	35.9	22.7	16.1
Group 3	4.5	3.1	2.3
From cancer surgeries			
Group1	47.5	11.3	8.4
Group2	22.4	26.8	19.9
Group 3	6.2	3.3	2.4

As shown in Table 2.1, the dielectric constants of normal breast tissues span a large range from 4 to 48. The results differ significantly from the small scale studies. It is suspected that there is disagreement in the definition of breast tissues used in the small scale studies and Lazebnik [48, 49]. The meaning of breast tissue assumed by the small scale studies is likely to be only adipose tissue in the breast, whereas Lazebnik [48, 49] considered adipose tissue, fibroconnective tissue, and glandular tissue in the breast.

In Lazebnik [48, 49], comparisons between cancer and normal breast tissues are made with three techniques. Technique 1 excludes samples from group 3 (mainly adipose tissue) and result shows the dielectric constant of cancer samples is only 8% higher than that of normal samples. Technique 2 includes only samples from group 1 (mainly glandular tissue) and result shows no difference between cancer and normal samples. Technique 3 includes samples from all groups and results show that the dielectric constant of cancer samples is 44.7% higher than normal samples. The new findings show that the dielectric contrast between cancer and normal breast tissues is much lower than the previously assumed 5:1 ratio.

The dielectric constant from previous assumption and Lazebnik [48, 49] are summarized in Table 2.2. The variability is calculated as 2 times of the standard deviation provided it does not exceed the range given by upper bound (saline) and lower bound (lipid) of the tissues dielectric constants. Therefore, the maximum variability of 100% calculated with 2 times of the standard deviation in Table 2.2, has to be replaced with range estimated to be 4 to 48 by referring to Table 2.1. Thus the variability for this special case is estimated to be $\pm 65\%$.

Table 2.2: Dielectric constants at 5 GHz of cancer and normal breast tissues from Lazebnik [48, 49] large scale studies and previously assumed values

	Mean	SE	SD	Variability
Lazebnik's study technique 1 (excludes adipose tissue)				
Cancer samples	49.8	1.17	7.1	$\pm 29\%$
Normal samples	46.1	1.80	9.5	$\pm 41\%$
Lazebnik's study technique 3 (includes all tissues)				
Cancer samples	49.0	1.09	7.6	$\pm 31\%$
Normal samples	33.8	1.83	16.9	$\pm 100\%$ (65%)
Previously assumed values				
Cancer tissues	50	—	—	0%
Normal tissues	9	—	—	$\pm 10\%$

For cancer tissues, there is no large discrepancy between small scale studies and Lazebnik [48, 49]. However for breast tissue dielectric constant, the mean value of 34 in Lazebnik [48, 49] is highly dependent on the sample collection procedure employed, where a sample with high adipose content and a sample with low adipose content are deliberately selected from each breast. The value can only represent the median between adipose and glandular tissues.

2.2 Microwave Breast Imaging

A comprehensive review of microwave breast imaging has been given by Fear et al [24]. The three basic approaches: passive, hybrid, and active are illustrated in Figure 2.2. .

Figure 2.2: Methods of microwave breast imaging. Figure taken from Fear et. al. [24]

Passive microwave imaging

Passive methods involve the use of radiometers to measure temperature differences in the breast, detecting tumors based on their increased temperature compared to normal tissue. Clinical results suggest that microwave radiometry has the potential to assist in the diagnosis of suspicious areas on mammograms.

Hybrid microwave-acoustic imaging

Hybrid methods use microwave energy to selectively and rapidly heat tumors due to the higher conductivity of malignant breast tissue. Ultrasound transducers are used to detect pressure waves generated by the expansion of the heated tissues.

Active microwave imaging

Active methods involve illuminating the breast with microwaves and then measuring the transmitted or reflected microwave signals. Classical approaches try to recover the complete dielectric properties profile of the breast. A more recent approach, namely confocal microwave imaging, seeks to identify the presence and location of significant scatterers in the breast.

The thesis focuses on UWB pulse-based confocal microwave imaging to locate the malignant breast tumors. As compared to the current gold standard, one disadvantage of UWB is that its resolution is not as high as mammography. However, UWB may provide better detection capability due to the larger contrast between tumor and normal breast tissues in microwave frequencies.

Some challenging problems of using UWB for breast cancer detection are difficulties in extracting the tumor response signals from the strong skin artifacts and differentiating the tumor response signals from the background clutter responses.

The skin artifacts problem is caused by the strong reflected signals from the breast skin which has dielectric constant higher than the breast tissues. The breast skin also causes attenuation to the incident signals and reflection signals from the breast interior. The problem is studied by researchers [31, 32, 33] and different approaches are proposed to suppress the skin artifacts. Other researchers focus on enhancing the tumor response signals out of the background clutter responses [18, 25, 26].

2.3 Algorithms for UWB based Breast Cancer Detection

2.3.1 Image Formation

There are several algorithms used for breast image formation using the backscatter signals. The simplest algorithm is delay-and-sum beamforming [13, 21], where signals are synthetically focused by phase shifting the signals to every location at the scanning region. An intensity image is created by summing the signals.

Microwave in Space Time Beamforming (MIST) [14, 15] algorithm has the same image formation concept as delay-and-sum beamforming. It is able to improve the signal-to-clutter ratio by compensating for frequency dependent propagation and attenuation by applying data adaptive filtering. A prior knowledge of tissue dielectric properties is needed to calculate the compensation factors.

There are several other algorithms which may improve the tumor response signal-to-noise ratio, such as, Generalized Likelihood Ratio Test [18], Data Adaptive Beamforming [25, 26], Time Reversal [27, 28] and Time of Arrival Data Fusion method [30].

2.3.2 Early Time Artifact Removal

There is a need to suppress early time artifact due to incident pulse and skin reflection. The best method is the subtraction algorithm which is always used as a benchmark for other algorithms. In subtraction method, signals from breast with tumor are subtracted with signals from breast without tumor to obtain the tumor responses. However the method is not applicable in practice but only in simulation studies for comparison purpose.

Another simple algorithm is averaging method [13, 21], where early time artifact is estimated from the averaged signals and removed from other antenna channels. Tumor responses can be recovered because they are not at the same position in time. However the method is not working well with realistic breast where the shape is not totally symmetrical.

There are several other algorithms which may be able to perform better early time artifact removal, such as, penalized least-squares method [15], Woody averaging and recursive least-square method [23], breast surface estimation [31, 32, 33], and entropy-based windowing method [34].

2.4 UWB Experiments for Breast Cancer Detection

Two research groups have done significant experiments on UWB for breast cancer detection, namely University of Wisconsin [16] and University of Calgary [23]. The experiments are conducted in frequency domain using a vector network analyzer to synthesize a pseudo UWB pulse. Breast phantoms are made of low loss dielectric medium with dielectric constant much lower than human breast tissue. The dielectric contrast between tumor and breast tissue is simulated instead of dielectric constants of human tissues.

University of Wisconsin: [16]

Research group in University of Wisconsin is led by Prof Susan C. Hagness. The equipment used is Agilent E8364A network analyzer with frequency range 10 MHz - 50 GHz. S_{11} parameter measurements are conducted for frequency range of 1 – 11 GHz for each antenna position where data is inverse fourier transformed to obtain the time domain backscattered signals.

Double-ridged pyramidal horn antennas with aperture of 2.5 x 2.0 cm are used. The antenna impedance is matched to the oil medium. The antenna array is planar rectangular array of 7 x 7 positions with 1 cm spacing.

Breast phantom is made of a large rectangular tank of oil to avoid reflection from boundaries. Tumor simulants are 4-mm-diameter x 4-mm-height plastic cylinder sealed with latex membrane containing diacetin solution. Solutions of different concentrations are used to simulate different dielectric contrasts relative to oil. A skin layer is created using 1.5-mm-thick FR4 glass epoxy printed circuit board. The materials used to construct the breast phantom are summarized in Table 2.3.

Table 2.3: Dielectric constant and conductivity of breast phantom materials at 6 GHz used by University of Wisconsin compared to human tissues

Human Tissue	ϵ_r	σ	Simulant	ϵ_r	σ
Breast tissue	9.8	0.4	Soy bean oil	2.6	0.05
Tumor	51	4.8	Diacetin solution	8.7	1.9
Skin	36	4.0	FR4 glass epoxy	4.3	0.016

Results with breast phantom images demonstrate the potential of the MIST [14, 15] algorithm to detect tumor. The minimum dynamic ranges required are 62 dB and 73 dB to detect tumors with dielectric contrast of 5.2 and 1.5 respectively, where dynamic range is defined as the source pulse voltage over tumor response voltage at a single antenna. Image signal-to-noise ratios (S/N) are 14.5 dB and 4.9 dB for tumors with dielectric contrasts of 5.2 and 1.5 respectively.

In the experiment, dielectric contrast is simulated instead of dielectric constants of human breast tissues. The reported dynamic range does not give an insight to what is required for UWB breast cancer detection prototype as the dielectric loss in oil is much lower than in breast tissues. The breast phantom is homogeneous without any clutters to represent the heterogeneous nature of human breast. The phantom is made so large to avoid multiple reflections from the boundaries. Measurement is done in frequency domain where much higher power is used.

Figure 2.3: Experimental setup for breast cancer detection at University of Wisconsin.
Figure taken from Xu Li et. al. [16]

University of Calgary: [23]

Research group in University of Calgary is led by Prof Elise C. Fear. The equipment used is Agilent 8719ES network analyzer with frequency range of 50 MHz – 13.5 GHz. S_{11} parameter measurements are conducted for frequency range of 1 – 10 GHz for each antenna position, data is inverse fourier transformed to obtain time domain backscattered signals.

The antenna used is Wu-King resistive monopole with length of 10.8 mm. Antenna array is a circular array of 16 positions with 22.5 degrees spacing. Breast phantom is made of a cylindrical tank of oil and flour mixture. Tumor is simulated by Alginate powder, water and salt mixture of 1 cm diameter. The materials used to construct the breast phantom are summarized in Table 2.4.

Table 2.4: Dielectric constant and conductivity of breast phantom materials at 4 GHz used by University of Calgary [23] as compared to human breast tissues

Human Tissue	ϵ_r	σ	Simulant	ϵ_r	σ
Breast tissue	9.9	0.4	Flour, oil, saline, mixture	4.2	0.16
Tumor	53	2.6	Alginate powder, water, salt, mixture	43.7	6.94
Skin	37	2.0	Epoxy sheet with dielectric fillers	34.3	4.25

Results with breast phantom images demonstrate the potential of confocal imaging algorithm to detect the tumor. Image signal-to-noise ratio (S/N) is 10.45 dB for 1 cm tumor.

The simulant which is claimed to have dielectric constant similar to human breast tissue is in fact much lower than the dielectric constant of human breast tissue. The dielectric contrast simulated is 10 times which is much higher than any published dielectric contrast between malignant and normal breast tissue. The performance of the monopole antenna used is poor.

Figure 2.4: Experimental setup for breast cancer detection at University of Calgary. Figure taken from Fear et. al. [23]

2.5 Breast Phantom Materials

In published experimental studies on UWB for breast cancer detection, soy bean oil is commonly used as the breast phantom material due to easy availability. Apart from oil, flour-oil-saline mixture [42], corn syrup and glycerine [43, 44] have been used as breast medium in UWB imaging experiments. However, these materials are unable to simulate the heterogeneous nature of human breast.

For building more realistic breast phantom which consists of low-water-content adipose tissues, high-water-content glandular tissues and cancerous lesions, the breast phantom materials are required to have the following features:

- Able to simulate the dielectric properties of human breast over the whole UWB frequencies range
- Able to exhibit heterogeneity and anthropomorphic configurations with long-term stability of mechanical and electromagnetic properties
- Simple fabrication process
- Raw materials used in fabrication are inexpensive and easily available

Tissue-mimicking phantom materials which simulate the properties of different biological tissues at discrete frequencies have been researched over the past decades [40, 41]. Recently, oil-in-gelatin dispersions with the above requirements have been proposed [39]. The materials can be created to have wide range of dielectric properties by varying the percentage of oil and gelatin.

The oil-in-gelatin dispersions have been used for ultrasound imaging studies 26 years ago [45, 47]. Anthropomorphic ultrasound breast phantom containing intermediate-sized scatterers has been created successfully [46]. For ultrasound application, the materials are made to mimic the acoustic impedances of the human tissues. For UWB imaging application, the same materials are used to mimic the dielectric properties of the human tissues.

Chapter 3: Methodology

This chapter describes the experimental setup and preliminary investigation done. Section 3.1 describes the overall experimental setup with introductions of the equipment used. Section 3.2 describes the algorithm used to process the recorded signals to produce a breast image showing the location of possible tumors. Finally, Section 3.3 describes the results of preliminary investigation using a simple breast phantom consists of soy bean oil.

3.1 Experimental setup

Figure 3.1: Overall Experimental Setup

The overall experimental setup is shown in Figure 3.1. UWB pulse is produced by Picosecond 3500D impulse generator, which is sent into the breast phantom by a Thales UWB antenna. Another Thales UWB antenna is used to receive the backscattered signals from the breast phantom. The signals are recorded by Agilent DSO81204B real-time oscilloscope and subsequently data is transferred to the computer. For frequency domain measurement, Agilent N5230A vector network analyzer is used in place of pulse generator and real-time oscilloscope.

The breast phantom is placed on an x-y-rotary stage with accuracy of 20 μm and 1 arcmin. A computer is used to control and synchronize the x-y-rotary stage movement and data transfer from the oscilloscope using Labview software. Figure 3.1 only shows antenna scanning in bistatic and circular configuration. Other variations of configuration are discussed in Section 3.1.2. Depending on the configuration, antenna is fixed at one position and the breast phantom is moved in x-y direction to simulate a planar antenna array or rotated for 360 degrees to simulate a circular antenna array.

3.1.1 Equipment used in Experiment

Picosecond Pulse Labs 3500D impulse generator as shown in Figure 3.2 is used to produce the UWB pulse. The output is gaussian pulse with full width at half maximum (FWHM) of 80 ps. The maximum pulse repetition rate is 1 MHz. Output power can be set by turning the attenuation setting from 0dB to 81dB, with maximum pulse amplitude of 8 V at 0dB. Delay time between trigger signal and output pulse can be set from 0 ns to 110 ns.

Figure 3.2: Picosecond 3500D Impulse Generator

Agilent DSO81204B real-time oscilloscope as shown in Figure 3.3 is used to record the signals received by the antenna. The bandwidth of the oscilloscope is 12 GHz with sampling rate of 40 GSa/s. The sampling rate can be extrapolated to 640 GSa/s with sinc(x) interpolation real-time digital processing. The vertical resolution is 16 bits, with vertical scale settings ranging from ± 1 mV to ± 5 V. The oscilloscope is able to perform real-time averaging up to 4096 waveforms. The dynamic range of the oscilloscope is 72dB with highest averaging setting.

Figure 3.3: Agilent DSO81204B Real-Time Oscilloscope

Agilent N5230A vector network analyzer as shown in Figure 3.4 is used in place of pulse generator and real-time oscilloscope for frequency domain measurement. Network analyzer functions as a sweep frequency generator and spectrum analyzer at the same time. Transmit and receive antennas are connected to the two ports of the network analyzer. The frequency range of the network analyzer is 300 kHz to 13.5 GHz. The maximum output power is 20 dBm whereas the minimum output power is -30 dBm. The dynamic range of the network analyzer is 110 dB.

Figure 3.4: Agilent N5230A Vector Network Analyzer

High precision x-y-rotary experiment stage as shown in Figure 3.5 is used to move the breast phantom relative to the stationary antenna to simulate a synthetic antenna array. The antenna is maintained stationary so that the channel multipath profile remains constant and the noise caused by cable movement is minimized. The linear accuracy of the stage is 20 μm whereas the rotational accuracy is 1 arcmin. Labview software is used to control the movement of the stage.

Figure 3.5: High precision x-y-rotary experiment stage

Thales UWB antennas as shown in Figure 3.6 are used as the transmitter and receiver of the UWB signals. The antennas dimension is 3cm width and 4cm height. The antennas gain is 11 dB with azimuth beamwidth of 60 degrees and elevation beamwidth of 40 degrees as shown in Figure 3.7. The antennas return loss measured with Agilent N5230A vector network analyzer is lower than -10 dB from 2.4 to 12 GHz as shown in Figure 3.8. Transmission is higher than -20dB from 1.8 to 6.3 GHz showing the 10dB bandwidth of nearly 5 GHz.

Figure 3.6: Front view and side view of Thales UWB antenna

Figure 3.7: Radiation pattern of Thales UWB antenna showing half-power beamwidth for azimuth of 60 degrees and elevation of 40 degrees.

Figure 3.8: S-parameters of Thales UWB antenna measured with two antennas facing each other separated by 5cm.

3.1.2 Planar Rectangular and Circular Antenna Arrays

In this section, two antenna array patterns are discussed and compared.

Planar rectangular array consists of antennas placed on top of the phantom as in the experiment conducted by University of Wisconsin. In a circular array, antennas are placed at the side near the wall of the phantom as shown in Figure 3. 9.

Circular array has an advantage where artifact can be easily removed by averaging method. This is because the multipath profile in the phantom remains the same during rotational scanning, unlike the planar rectangular array where the phantom boundary changes when antenna position changes. Circular array also provides better illumination on the tumor from all directions and gives better summed tumor response.

Figure 3.9: Experiment setup for circular antenna array

3.2 Signal Processing

This section describes the delay-and-sum algorithm for confocal microwave breast imaging commonly employed in breast cancer detection. An image of tumor is obtained by synthetically focusing the signals received from an antenna array. The flow chart of the signal processing steps is shown in figure 3.10. Detailed description of each step and the signals obtained are shown in figure 3.11.

The simple delay-and-sum algorithm is used for the calibration of experimental setup for tumor detection using simple homogenous breast phantoms. Modifications to compensate for hardware limitations will be made for more realistic breast phantoms to be discussed in Chapter 4.

Figure 3.10: Flow chart of Delay-and-Sum algorithm.

Figure 3.11a: Raw received signals

All recorded raw backscattered signals in Figure 3.11a are normalized to 1 for artifact removal at the next step.

Figure 3.11b: Artifacts removal

Artifacts include the incident pulse, phantom boundary reflection, and multipath due to the phantom structure. An artifact template is created by averaging the signals from all antenna positions. Due to high similarity of artifacts in all the signals, tumor responses can be obtained by subtracting the averaged signal from each received signals.

Figure 3.11c: Phase shifting of signals after delay time estimation

Delay time is estimated from the distance of the antenna to every location in the scanning region and the speed of light in the breast phantom. For each location, all signals are phase shifted accordingly. Aligned signals are tumor responses as shown in Figure 3.11c.

Figure 3.11d: Loss compensation

Attenuation of signals is caused by distance effect of the radiating pulse and loss due to the medium dielectric properties. Loss compensation is done by multiplying the corresponding loss factors such that all the tumor responses have equal amplitude as shown in Figure 3.11d.

Figure 3.11e: Summing and windowing of signals

All signals are summed and windowed for energy calculation. Note that the signals add coherently for the tumor responses but add non-coherently for non tumor responses. Grey solid line is early time artifact for reference purpose to set the window position.

Figure 3.12: Energy calculation and image plotting

The energy of the windowed signal is calculated for every location and plotted in two dimensions for that breast slice. Bright spot in Figure 3.12 indicates the tumor location. Note that the image is a plot showing the location of possible UWB pulse scatterers. It does not represent the structure and dielectric properties of the breast. The size of the bright spot is dependent on the image resolution which is related to the speed of light in the breast phantom and the pulse width used. It does not indicate the size of the tumor.

3.3 Calibration experiments

The calibration experiments are conducted using a simple homogenous breast phantom made of a cylindrical polypropylene container filled with soy bean oil to test the detectability of tumor. Soy bean oil as shown in Figure 3.13 is used because of easy availability. Other cooking oil should have similar dielectric properties.

Figure 3.13: Simple breast phantom of oil (Left) and small screws used as tumor simulants (Right)

The breast phantom is a cylindrical polypropylene container (diameter = 16cm, height = 12cm) filled with soya bean oil. Tumor simulants (small metal screws with diameter = 2mm, length = 5mm) are suspended in oil with polyester thread. For calibration purpose, metal items are used as ideal targets because they are perfect reflectors at microwave frequencies.

An antenna pair is placed at the side of the phantom close to the container wall as shown in Figure 3.9. The antenna pair is scanned around the phantom by rotating the phantom for 360 degrees to simulate a circular array of 24 antennas with 15 degrees spacing. Signals collected are processed with delay-and-sum algorithm as discussed in section 3.2. Images are then formed with resolution of 1 cm^2 within the $17 \text{ cm} \times 17 \text{ cm}$ region.

Results obtained are shown in Figure 3.14. Grey color images are created for reference purposes to show where the tumors are placed in the experiments, as indicated by the black pixels. The images formed are in black color background with tumors indicated with white pixels.

Figure 3.14a: Imaging results with one tumor simulant.

Figure 3.14b: Imaging results with two tumors simulants.

Figure 3.14c: Imaging results with two close tumor simulants

Chapter 4:

Experimental Issues and Compensation Methods

Recent researches based on numerical modeling which ignored hardware characteristics and simple experiments using homogenous breast phantoms have shown the potential of ultra-wideband to detect early stage breast cancer. However, clutter interference from heterogeneous breast tissues and hardware characteristics like pulse instability, finite dynamic range and precision for signal acquisition can severely degrade the detectability of breast tumors. This chapter discusses the experimental issues encountered and compensation methods used to improve the detectability of tumor.

In order to bridge the gap between numerical simulations and experiments, it is important to identify the experimental issues before conducting experiments with more realistic breast phantoms so that the source of imaging artifacts can be identified and compensated. An ideal imaging scenario is first created where the simple sum-and-delay algorithm is working perfectly. In this ideal scenario, the breast phantom is round and symmetrical such that the averaging method can perfectly remove the early time artifact. The breast medium is homogeneous such that propagation of signal in the medium is constant with accurate delay time estimation. Skin is approximated by a single interface (air to breast phantom) because its thickness is small compared to UWB pulse width in space.

Several factors that give rise to imaging artifacts are:-

- Equipment noise caused by pulse instability of the UWB pulse generator and limited dynamic range of the real-time oscilloscope.
- Environmental noises captured by the UWB antenna, such as 1.8GHz and 2.4GHz interferences due to wireless communications.
- Positioning errors of the breast phantoms, non-uniform distribution of dielectric properties, and imperfect shape of the breast phantoms.

Section 4.1 discusses the pulse instability imaging artifacts and compensation. Power loss compensation is considered in section 4.2. Filtering and correlation applied before delay-and-sum beamforming algorithm are shown to improve the image SNR in section 4.3. Section 4.4 discusses the ring artifact caused by positioning error of breast phantom. The dynamic range of the real-time oscilloscope used in the experiment is investigated in section 4.5. Finally, section 4.6 considers the effects of averaging settings and number of antennas used on image SNR.

4.1 Pulse Instability Artifacts and Compensation

As discussed in the previous chapter, averaging method is applied to remove artifacts in the received signals before delay-and-sum beamforming. An averaged signal of all received signals from different antennas is calculated. The averaged signal is used as a template artifact and is subtracted from individual received signals. Clean tumor responses can be obtained if the system is free of noise.

However, averaging method will not work well if the received signals are not aligned perfectly and have unequal amplitudes. Pulse phase instability is caused by the impulse generator being unable to maintain a constant delay time between trigger signal and the output UWB pulse. Pulse amplitude instability is caused by the impulse generator which is unable to maintain constant amplitude of the output UWB pulse. Pulse instabilities have not been considered by UWB breast cancer detection researchers.

Figure 4.1a and 4.2a show that pulse instability causes phase shift of 20 sample points or 31 ps in the signals. As shown in Figure 4.3a, the resultant noise in the subtracted signals is large. In phase instability compensation, all received signals are aligned by finding the zero-crossing point between maximum and minimum peaks and phase shifting the signals to the same zero-crossing point. For amplitude instability, compensation is done by normalizing all the signals peak-to-peak amplitude to one.

Figure 4.1a: Signals before pulse instability compensation. **Figure 4.1b, 4.1c, and 4.1d:** Signals after pulse instability compensation at sampling rate of 40GHz and extrapolated sampling rates of 640GHz and 3THz.

Figure 4.2: Zoom in view of signals given in Figure 4.1.

Figure 4.3a: Resultant noise after averaging method for signals without pulse instability compensation. **Figure 4.3b, 4.3c, 4.3d:** Resultant noise after averaging for signals with pulse instability compensation at 40GHz, extrapolated 640GHz and 3THz.

In order to further minimize the phase error, the received signals are extrapolated to higher sampling rates. Table 4.1 shows the calculated worst phase error for received signals and resultant noise after pulse instability compensation at different extrapolated sampling rates. Calculations are done in Matlab by subtracting two signals, one signal is phase shifted by one sample time from another signal.

Table 4.1: Calculated and measured noise amplitude after pulse instability compensation at different extrapolated sampling rates.

Sampling Rate	Worst Phase Error	Calculated Noise Amplitude	Measured Noise Amplitude
No compensation	31.25 ps	–	0.7866
40 GHz	25.00 ps	0.5437	0.5217
640 GHz	1.56 ps	0.0533	0.0565
1 THz	1.00 ps	0.0344	0.0436
2 THz	0.50 ps	0.0173	0.0435
3 THz	0.33 ps	0.0117	0.0437
4 THz	0.25 ps	0.0088	0.0437
5 THz	0.20 ps	0.0071	0.0434
10 THz	0.10 ps	0.0036	0.0433

The noise amplitude shown is relative to the incident pulse amplitude. Table 4.1 shows that the measured resultant noise amplitude does not further decrease with higher sampling rates greater than 1 THz because the other contributors of noise such as environmental noise starts becoming significant.

Figure 4.4 shows the effects of pulse instability compensation on images formed by delay-and-sum beamforming. The tumor is located at 3 cm to the right from the center.

Figure 4.4: The effects of pulse instability compensation on breast phantom images. Left: without compensation. Right: compensation at extrapolated 640GHz.

4.2 Loss Compensation

This section discusses the three factors contributing to power loss during propagation of UWB pulse in breast medium, and the loss models used for compensation. Loss compensation is a signal processing step to equalize all the received signals originating from different locations. There are three loss factors:

- Radial spreading
- Antenna pattern
- Medium dielectric

In most simulation studies, only the first factor is considered and the remaining two are ignored. Thus the signal loss compensation is commonly mentioned as radial spreading compensation.

4.2.1 Radial Spreading Loss

Most studies approximate the propagating signal as a uniform cylindrical wave and thus the radial spreading loss is proportional to distance. Considering both transmit and receive paths make the loss proportional to distance square. Compensation is done by multiplying the signals by r^2 , where r is distance from antenna to the particular scanning point. Figure 4.5 shows the decrease of reflected signal amplitude from a tumor considering only radial spreading loss. The tumor is located nearest to the antenna at 0 degree and furthest at 180 degrees.

Figure 4.5: Simulated received pulse amplitude considering only radial spreading loss

4.2.2 Antenna Pattern Loss

Antenna pattern loss is caused by non-uniform gain of the directional antenna used. The Thales UWB antenna used in our experiment has half-power beamwidth of 60 degrees as shown in Figure 4.6.

Figure 4.6: Normalized gain of Thales antenna with half-power beamwidth of 60 degrees

Figure 4.7: Antenna gain in the scanning region for transmit and receive antennas

Antenna pattern loss causes the decrease in reflected signal power when the scanning point is not within beamwidth of the antenna. Simulated signal amplitude as shown in Figure 4.8 is obtained by multiplying the transmitter gain and receiver gain as shown in Figure 4.7. The tumor is located nearest to the antenna at 0 degree and furthest at 180 degrees.

Figure 4.8: Simulated received pulse amplitude considering radial spreading loss and antenna pattern loss

4.2.3 Medium Dielectric Loss

Medium dielectric loss is related to the dielectric properties of the medium. For low loss medium (loss tangent $\epsilon''/\epsilon' \ll 1$) the signal amplitude, A , and attenuation constant α , is given by:

$$A = e^{-\alpha x} \quad \alpha = \frac{\sigma}{2} \sqrt{\frac{\mu}{\epsilon}}$$

where x is the wave travel distance, σ is conductivity, μ is permeability, ϵ is permittivity. At UWB frequencies, breast tissues can be considered as low loss medium. By referring to Table 4.2, the signal amplitude will be half at 3.5 cm depth of breast tissues. For UWB pulse, this is an approximation using a single frequency without considering the effect of dispersion.

Table 4.2: Alpha and Attenuation for different mediums at 5 GHz

Medium	ϵ''/ϵ'	σ	α (cm ⁻¹)	A (at 1 cm)
Oil	0.079	0.057	0.067	0.935
Breast	0.12	0.33	0.20	0.82
Water	0.28	5.71	1.25	0.29

4.2.4 Effects of Loss Compensation

To see the effects of loss compensation, compensation is applied on experiment data to compare the results obtained without compensation applied.

Figure 4.9: Image with radial spreading compensation (left) and without radial spreading compensation (right).

Imaging results in Figure 4.9 show that loss compensation tends to amplify noise near the phantom boundary. The compensation applied here is only considering radial spreading loss. Worse results will be expected if other loss factors are incorporated, since the signals are multiplied by larger factors.

Radial spreading compensation is a commonly used signal processing step in beamforming algorithms in numerical noise-free modeling. In view of the deteriorating effects of radial spreading compensation on image quality, it is recommended not to apply the compensation.

4.3 Filtering and Correlation

This section describes two signal processing methods applied to improve the signal-to-noise ratio (SNR) of the breast phantom images.

4.3.1: Filtering

The ultra-wideband (UWB) antenna used in the experiments has a bandwidth of 1.8 to 6.3 GHz. Two significant narrowband interferences for the experiments conducted are cell phone noise at 1.8 GHz and wireless local area network (LAN) noise at 2.4 GHz. Digital notch filters at 1.8 GHz and 2.4 GHz are applied to reduce the interferences. The signals and power spectrums before and after filtering applied are given in Figure 4.10. The image quality has been improved with filtering as shown in Figure 4.11.

Figure 4.10a: Signals and power spectrums before filtering.

Figure 4.10b: Signals and power spectrums after filtering.

Figure 4.11: Images formed without filtering (left) and with filtering applied (right). The tumor is located at 3 cm to the right of the center.

4.3.2: Correlation

A tumor response template has been obtained in the calibration experiments described in Chapter 3 as shown in Figure 4.12. Correlation is applied by multiplying the tumor response with the filtered windowed signals after the delay-and-sum operation at each pixel. Then the filtered windowed signals are summed to give a value for every pixel in the breast phantom image given in Figure 4.13. The image quality has been enhanced with correlation.

Figure 4.12: Tumor response template

Figure 4.13: Comparison for filtered images without correlation (left) and with correlation (right)

All the discussion above focuses on only a single experiment data. The result may not be convincing because occurrences of artifacts in images are random due to coherence alignment of noise at certain points. Thus large scale experiments are performed to study the effect of applying filtering and correlation on breast phantom images. Ten experiments are conducted with the same experimental setup and breast phantom. Tumor simulant is at position (131, 101) for all experiments. Experiments are repeated with different (6, 12, 24, and 36) antennas array in one phantom rotation. Forty sets of data are collected and processed. A total of 80 images are formed for SNR analysis. Figure 4.14 show 20 images from experiments conducted with 24 antennas array.

Figure 4.14a: Five filtered images without correlation (top) and corresponding filtered images with correlation (bottom)

Figure 4.14b: Five filtered images without correlation (top) and corresponding filtered images with correlation (bottom)

SNR is calculated from the ratio of the pixel value at tumor location (131,101) over the highest value of clutter pixels 6 mm radius outside the tumor location. Since the SNR value from individual experiment is highly variable, due to random occurrences of artifacts, an averaged SNR value is calculated using ten SNR values from ten experiments.

Detectability is the ability to observe the tumor in the images although the tumor may not appear as the strongest pixel. Detectability is defined as one when the tumor pixel value is above half of the maximum scale and above twice of the adjacent region pixel values, otherwise zero is given for that image. Table 4.3 shows that detectability is at least 80% when SNR is positive.

Table 4.3: SNR and detectability for different antennas arrays

SNR	6 antennas	12 antennas	24 antennas	36 antennas
Without Correlation	-6.3 dB	0.5 dB	6.1 dB	8.2 dB
Correlation	-5.7 dB	3.5 dB	11 dB	13 dB
Detectability	6 antennas	12 antennas	24 antennas	36 antennas
Without Correlation	1/10	8/10	10/10	10/10
Correlation	2/10	10/10	10/10	10/10

Figure 4.15 show the SNR versus number of antennas per phantom. There is significant improvement on SNR with correlation applied. Increasing the number of antennas improves the SNR but cost and complexity of implementation increases. The minimum number of antennas needed without correlation is 12 compared to 10 for correlation.

Figure 4.15: SNR versus number of antennas

In order to show the robustness of the correlation method, the tumor stimulant is positioned at different distances, 1 cm to 6 cm, from center of the breast phantom with images formed given in Figure 4.16. Images in Figure 4.16 show that correlation improves the image quality.

Figure 4.16: Images formed without correlation (top) and with correlation (bottom), with tumor located 1 cm to 6 cm from center.

4.4 Ring Artifact

In some experiments with higher dielectric constant phantoms, ring artifacts appear as shown in Figure 4.17. The images shown in Figure 4.17 are obtained from an experiment using a high dielectric phantom.

The phantom is fabricated using tissue mimicking phantom material [39] with 80% oil, contained in a polypropylene cylindrical container (diameter 16cm, height 12cm). The material is able to closely simulate the dielectric properties of human tissues. Tumor simulant is an 8 mm cube made of phantom material with 10% oil buried at 25 mm from center of the phantom. The measured dielectric constant at 5 GHz for phantom medium is 9 whereas for tumor is 50 which is representative of normal and malignant human breast tissues. Ring artifacts have not been reported by other researchers because their breast phantoms use lower dielectric constant materials.

Figure 4.17a: Images of breast phantom 1 mm off-center (left) and breast phantom less than 0.5 mm off-center (right)

Ring artifact arises from positioning error (off-center) of the breast phantom. Ideally, averaging method works perfectly with round phantom. Ring artifact appears when the phantom is not positioned perfectly on the rotary axis of the experimental stage. This causes small displacement error of the phantom boundary relative to the antennas.

Ring artifact is caused by the coherent adding of the residue boundary reflections after delay-and-sum beamforming. The ring-to-ring distance is proportional to the wavelength of the incident signal. Ring artifact also indicates the direction of phantom off-center displacement. For instant, when the phantom is displaced to right side, the ring artifact will appear on right indicating positive x-axis off-center displacement and left indicating negative x-axis off-center displacement.

In experiments with oil medium which has lower dielectric constant, ring artifact is not noticeable because the tumor response is large enough to dominate the ring artifacts in the image. The correlation method is not able to improve the image quality when ring artifacts arise as illustrated in Figure 4.17b. This is due to the high similarity between tumor response and the residue incident pulse ringing.

Figure 4.17b: Same Images from Figure 4.17a with correlation applied

Adjusting the phantom to the best position using visual inspection will result in a positioning error of 1 mm to 3 mm. Better placement can be achieved by placing a reference object on the antenna to measure the antenna to phantom boundary distance and adjusting the phantom position until the error is smaller than 0.5 mm. The resulting ring artifact is reduced significantly.

4.4.1 Experiment on Ring Artifact

An experiment is conducted to test the amplitude of the ring artifact for different displacement errors of a breast phantom without tumor.

A small container (diameter 10 cm) of oil is used as breast phantom without tumor. The phantom is adjusted to the best position with error less than 0.5 mm. Measurements are taken for the phantom at best position, then with off-center displacements of 1 mm, 2 mm, and 3 mm from the best position. The resultant noise shown in Table 4.4 is computed after applying averaging method. The result shows small off-center displacements can contribute to significant noise as illustrated in Figure 4.18 and subsequently cause ring artifacts as shown in Figure 4.19.

Table 4.4: Averaged RMS and peak-to-peak amplitude of resultant noise after averaging method for different off-center positioning errors

Position error	Averaged Noise RMS ($\times 10^{-3}$)	Averaged Noise P-P ($\times 10^{-3}$)
<0.5 mm	1.0	2.3
1 mm	3.3	5.7
2 mm	5.5	9.9
3 mm	8.7	15.0

Figure 4.18: Resultant noise after averaging method for best positioning (top) and 1mm displacement from center (bottom). Noise is three times larger with only 1 mm error.

Figure 4.19a: Images for phantom with best positioning (left) and 1mm off-center displacement (right)

Figure 4.19b: Images for phantom with 2mm (left) and 3mm (right) off-center displacement. Note that the pattern does not change but the amplitude changes.

4.5 Dynamic Range of Real-Time Oscilloscope

This section discusses the dynamic range of Agilent DSO-081204B real-time oscilloscope used in the experiment. In this discussion, dynamic range is defined as the ratio of vertical scale range and smallest amplitude of digitized pulse. This differs from the definition in breast cancer detection system, whereas dynamic range is defined as the ratio of incident pulse amplitude to the tumor response amplitude.

Vertical resolution of the oscilloscope is first determined. The dynamic range can be estimated 4 or 5 bits smaller than the vertical resolution, as the bits are required to construct the smallest pulse.

4.5.1 Vertical Resolution of Real-Time Oscilloscope

Agilent DSO-081204B real time oscilloscope has front ADCs with 8-bit resolution. After real time averaging and interpolation, the oscilloscope is able to increase the vertical resolution and save the data in 16-bit format.

In order to determine the actual vertical resolution of the saved data at different averaging settings and different vertical scale settings, a histogram of the saved data is plotted with very small divisions (65536 divisions) so that every discrete value can be separated as shown in Figure 4.20.

For continuous signal, the histogram bars will be adjoining each other. As for discrete signal, the histogram bar will be separated. The number of bars for a range of value is calculated to determine the vertical resolution.

$$\text{Bits of resolution} = \frac{\ln \left[\left(\frac{b}{r} \right) \times s \times 8 \right]}{\ln(2)}$$

where, b is number of bars

r is range of value

s is vertical scale range

Figure 4.20: Histogram of a pulse recorded from the Agilent real-time oscilloscope.

Table 4.5: Bits of resolution for different averaging settings and vertical scales

Scale	1avg	2avg	4avg	8avg	16avg	32avg	64avg
1mV	13.0	13.0	13.0	13.0	13.8	13.9	13.9
2mV	10.6	10.6	10.6	10.6	14.0	14.0	14.0
5mV	12.0	12.0	12.0	12.0	15.3	15.3	15.3
10mV	13.0	13.0	13.0	13.0	16.0	16.0	16.0
20mV	10.6	10.6	10.6	10.6	14.0	14.0	14.0
50mV	12.0	12.0	12.0	12.0	15.3	15.3	15.3
100mV	13.0	13.0	13.0	13.0	16.0	16.0	16.0
200mV	10.6	10.6	10.6	10.6	14.0	14.0	14.0
500mV	12.0	12.0	12.0	12.0	15.3	15.3	15.3
1000mV	13.0	13.0	13.0	13.0	16.0	16.0	16.0

For higher averaging settings, 128, 256, 512, 1024, 2048, 4096, the vertical resolutions are same as 64avg. Table 4.5 shows that the maximum vertical resolutions occur when the vertical scale is set to 10mV, 100mV, or 1V. This provides a guideline to set the pulse amplitude to fully occupy the vertical range which is 80mV, 800mV, or 8V, so that the vertical resolution is fully utilized.

4.5.2 Dynamic Range of Real-Time Oscilloscope

The maximum vertical resolution provided by the oscilloscope is 16 bits. Due to quantization of the signal, at least 4-bit resolution is needed to construct the pulse shape of tumor response as shown in Figure 4.21. So the remaining 12-bit range is the maximum dynamic range of the oscilloscope.

Figure 4.21: Ideal tumor response constructed with 2-bit, 4-bit, and 6-bit vertical resolutions showing that the minimum resolution required is 4 bits.

A simple experiment is conducted to determine the dynamic range of the oscilloscope by looking at the ability to construct a pulse. Due to the presence of noise, a pulse cannot be observed from a single signal. Instead, 360 signals are collected and an averaged signal is obtained. The random noise will be averaged out, and pulse can be observed if the pulse is within the dynamic range.

The incident pulse is set so that its amplitude is equal to the expected smallest peak for different possible dynamic range as shown in the Table 4.6. The recorded signals are averaged to determine at which dynamic range the pulse can still be constructed.

When the pulse amplitude is set to 2mV, the pulse cannot be seen from the averaged signals, whereas for 4mV, the pulse is merely noticeable as shown in Figure 4.22. From this experiment which considers the system noise, the dynamic range is estimated to be 11-bits, which has 2048 values available for the whole range. Thus, the maximum acceptable ratio between incident pulse and tumor response is 2048.

Table 4.6: Expected and recorded peak of incident pulse for dynamic range

Dynamic Range $2^{\wedge}$	Expected smallest peak	Recorded peak	Averaged peak
8bits	31mV	33.3mV	33.2mV
9bits	16mV	17.7mV	14.9mV
10bits	8mV	11.7mV	7.8mV
11bits	4mV	9.8mV	6.1mV
12bits	2mV	8.9mV	5.9mV

$$\text{Expected peak} = \frac{\text{Full range voltage}}{\text{Dynamic range}} = \frac{8V}{2^{\wedge}12\text{-bits}} = 0.002V$$

Figure 4.22: Constructed pulse to test dynamic range of $2^{\wedge}11$ -bits. The pulse is just noticeable at sample 1600.

4.6 Signal and Image SNR

This section discusses the effects of improving the image SNR by increasing the averaging number and antenna number. The tradeoff of increasing these two factors is the increase of acquisition time.

In most simulation studies, array of only a few antennas are considered. The reasons are more antennas do not increase the image resolution, and do increase the simulation time. However, the advantage of more antennas is noise reduction.

Better image SNR can be obtained by increasing the averaging number and antenna number. This section investigates the optimum averaging number and antenna number to be used for the experimental setup.

Some terms used in the discussion are defined as follows:

- Averaging number is the number of waveforms acquired by the oscilloscope to produce an averaged waveform for each acquisition at one antenna position.
- Antenna number is the number of antennas in the synthetic array around the circumference of the phantom. It is also the number of steps for a complete phantom rotation relative to a stationary antenna.
- Signal SNR is the ratio of the root mean square (RMS) value of averaged tumor response to the RMS value of the averaged noise.
- Image SNR is the ratio of the tumor pixel intensity to the highest-value artifact pixel intensity. The definition differs from SCR which is Signal-to-Clutter Ratio, since there is no clutter considered in this discussion. Artifact pixels are caused by coherent summation of noise and occur randomly whereas clutters remain at their positions and averaging is unable to remove them.

4.6.1 SNR Improvement with Larger Averaging and Antenna Number

Signal SNR can be improved by using larger averaging number, whereas image SNR can be improved by using larger antenna number.

To measure the noise level, antenna is placed stationary without the presence of breast phantom or any moving object. A total of 360 measurements are taken as in breast scanning. Noise is calculated by subtracting the individual measurement with the average measurements.

The noise in the pulse is calculated instead of directly measuring the noise without the incident pulse. The RMS and peak-to-peak noise as shown in Table 4.7 is the average of 360 noise amplitudes normalized to the incident pulse amplitude.

Table 4.7: RMS and peak-to-peak amplitude of noise for different averaging number with maximum pulse amplitude of 8V (attenuation setting 0 dB)

Averaging No.	RMS noise	P-P noise
1	4.8 mV	18.4 mV
2	2.3 mV	10.2 mV
4	1.7 mV	7.1 mV
8	1.2 mV	4.9 mV
16	0.83 mV	3.6 mV
32	0.57 mV	2.49 mV
64	0.45 mV	1.71 mV
128	0.33 mV	1.35 mV
256	0.24 mV	0.99 mV
512	0.20 mV	0.82 mV
1024	0.15 mV	0.67 mV
2048	0.12 mV	0.54 mV
4096	0.10 mV	0.49 mV

Figure 4.23: RMS noise versus averaging number for different pulse attenuation settings

Matlab simulations are conducted to determine the image SNR. Ten sets of UWB noise are created by applying a bandpass filter to white noise with all the resultant noise RMS amplitudes set to 0.5 mV. The noise is scaled to the noise RMS amplitude of different averaging numbers as shown in Table 4.7 and added to the ideal tumor response as shown in Figure 4.12. The tumor response has peak-to-peak amplitude of 1 mV and RMS of 0.2567 mV.

The signals are delayed as though they are received from the 360 antennas spaced regularly around the breast phantom. The same delay factors are used in creating the signals and in subsequent beamforming. Thus there is no error caused by delay estimation. A total of 120 sets of data are created for confocal imaging for each antenna number of 12, 24, 45, 90, 180, and 360.

Since the image SNR is random due to random occurrences of artifacts, ten SNR values are obtained and averaged based on simulation with ten sets of data for each averaging number and antenna number. A total of 720 images are produced from the simulations. The results are shown in Table 4.8.

Images from the simulations are shown in Figure 4.24, 4.25, and 4.26 for different sets of noise, noise amplitudes, and antenna numbers.

Table 4.8: Image SNR for different averaging number and antenna number. Signal SNR is calculated with tumor response RMS of 0.2567 mV.

Antenna No.		12	24	45	90	180	360
Averaging No.	Signal SNR	Image SNR					
2	-18.66	-2.8	-2.8	-2.2	-2.4	-2.0	-1.3
4	-15.89	-2.8	-2.8	-2.1	-2.1	-1.4	0.0
8	-13.4	-2.7	-2.7	-1.9	-1.7	-0.2	1.7
16	-10.09	-2.6	-2.2	-1.3	-0.2	1.9	4.0
32	-7.08	-2.3	-1.3	-0.2	1.9	4.0	5.8
64	-4.68	-1.9	-0.2	1.3	3.8	5.5	6.8
128	-2.69	-1.2	1.1	2.6	5.1	6.5	7.5
256	-0.11	0.3	2.8	4.4	6.4	7.4	8.1
512	2.17	1.9	4.3	5.7	7.2	8.0	8.6
1024	3.58	2.9	5.2	6.4	7.7	8.3	8.8
2048	6.6	4.8	6.5	7.5	8.4	8.8	9.1
4096	8.19	5.6	7.1	8.0	8.7	9.0	9.2

Figure 4.24: Image SNR versus Signal SNR for different antenna numbers.

As observed from Figure 4.24, the cost to increase the image SNR is getting higher when the tumor responses SNR is positive. When the tumor responses SNR is negative, the cost to increase the image SNR is lower. In other word, it is worth using array of more antennas when the tumor response received is weaker than the noise.

In this simulation, the amplitude of the tumor response is fixed at 1 mV, with RMS value of 0.2567 mV. The tumor response is 1000 times smaller than the 1V incident pulse. In this case, we could use antenna number of 12 and averaging number of 1024 to produce reasonably good results.

Figure 4.25: Images obtained with ten sets of noise processed with antenna number of 360 and RMS noise level of 2.2 mV

Figure 4.26: Images obtained with the same set of noise processed with antenna number of 360 and RMS noise level from 2.2 mV (top left) to 0.10 mV (bottom right)

Figure 4.27: Images obtained with the same set of noise processed with antenna number of 12 (top left), 24, 45, 90, 180, 360 (bottom right) and RMS noise level of 0.45 mV

Chapter 5:

Fabrication and Dielectric Measurement of Breast Phantoms

This chapter describes the fabrication and measurement of dielectric properties of breast phantom materials using Agilent N5230A vector network analyzer. Measurements are conducted on material surface using open-ended coaxial probe. Reflection coefficients are converted to dielectric constants and loss tangents by Agilent 85070 dielectric measurement software.

Section 5.1 describes the probe performance and verification using standard materials. The time effects on measurement accuracy are considered in section 5.2. Section 5.3 and 5.4 discusses the measurement of dielectric properties of phantom materials fabricated in cylindrical containers and flat rectangular containers respectively. Section 5.5 describes the fabrication of homogeneous and heterogeneous breast phantoms for breast cancer detection experiments using the phantom materials. Section 5.6 and 5.7 discussed the dielectric properties of homogeneous and heterogeneous breast phantoms fabricated in cylindrical containers. Section 5.8 discussed the analysis of dielectric variability of breast phantoms fabricated.

5.1 Probe Performance

The Agilent 85070 slim-formed open-ended coaxial probe is used for dielectric properties measurement as shown in Figure 5.1. The measurement accuracy and repeatability specified by the manufacturer are 5% and 1.5% respectively.

The specification is verified using standard materials with well known dielectric properties. The materials used are distilled water, methanol, ethanol, and propanol. 64 measurements are taken with calibration conducted after every 16 measurements.

Figure 5.1: Open-ended coaxial probe used for dielectric properties measurement

Figure 5.2: Dielectric constant and conductivity of water measured with Agilent probe.

The measurements of dielectric constant and conductivity for distilled water are shown in Figure 5.2. Measurements are repeated for three Agilent probes available. The measurement error and range are shown in Table 5.1. The standard values are calculated with Debye equation:

$$\epsilon'_r + j\epsilon''_r = \epsilon_\infty + \frac{\epsilon_s - \epsilon_\infty}{1 + j\omega\tau} - j \frac{\sigma_s}{\omega\epsilon_0}$$

where ϵ_r is the relative complex permittivity, ϵ_∞ is the permittivity at high frequency, ϵ_s is the permittivity at low frequency, σ_s is the static conductivity, $\epsilon_0 = 8.85 \times 10^{-12}$ is the permittivity at free space, and ω is the angular frequency. The parameters for standard materials are listed in Table 5.2. In order to verify the accuracy specified by the manufacturer, standard materials with well known dielectric properties are used.

Table 5.1: Water dielectric constant at 0.5 GHz for three coaxial probes

	Debye	Mean	Error	Range
Probe-1	80.33	80.04	0.29 (0.4%)	2.37 (3.0%)
Probe-2	80.33	80.08	0.25 (0.3%)	2.51 (3.1%)
Probe-3	80.33	79.84	0.49 (0.6%)	2.19 (2.7%)

Table 5.2: Debye parameters for standard materials [35]

	ϵ_{∞}	ϵ_s	τ
water	5.2	80.4	9.45×10^{-12}
methanol	5.7	33.6	53×10^{-12}
ethanol	4.2	25.1	143×10^{-12}
propanol	3.5	19.5	332×10^{-12}

Figure 5.3: Dielectric constants and conductivities of methanol, ethanol, and propanol

Table 5.3: Dielectric constants for three standard materials at 5 GHz

	Debye	Mean	Error	Range
methanol	13.11	12.80	0.31 (2.4%)	0.05 (0.4%)
ethanol	5.18	5.10	0.08 (1.5%)	0.04 (0.7%)
propanol	3.65	3.78	0.13 (3.6%)	0.04 (1.0%)

The probe accuracy is verified from the measurement error using the three standard materials as shown in Figure 5.3. The measurement error is calculated from the mean of 64 measurements compared to the value calculated with Debye equation. The range is calculated from the minimum and maximum measurement values.

For dielectric constant as shown in Table 5.3, the errors for three standard materials are within the specified accuracy which is 5%, whereas the ranges are within the specified repeatability which is 1.5%.

5.2 Time Effect on Measurement Accuracy

In order to improve the measurement accuracy, time effects on measurement result is studied. The first effect is due to the time of measurement being conducted after the calibration of probe is done. The second effect is the time of measurement being conducted after the phantom material is taken out from the container.

To study the first effect, two measurements are taken using distilled water one hour after the probe calibration is done. One measurement is conducted with electronic calibration auto-refresh function turned on, and another measurement is conducted without the electronic calibration auto-refresh function turned on.

It is found that the measured water dielectric constant changes when the auto-refresh function is turned off. The effect of turning on and off the auto-refresh function is shown in Figure 5.4. Thus, electronic calibration auto-refresh function should be turned on to ensure reliable measurement over time.

Figure 5.4: Water dielectric measurements where electronic calibration auto-refresh function is turned on and off. Measurement performed 1 hour after calibration.

The second effect is due to the time of measurement being conducted after the phantom material is taken out from the container. During measurement, the phantom material is exposed to air. It is observed that the phantom material dielectric constant will decrease over time due to dehydration.

To study this effect, a simple experiment is conducted to estimate the rate of decrease. A small phantom material of 1 cm cube is used for the experiment. The experiment is conducted within 12 hours with the cube exposed to air. Measurement is conducted every half an hour.

Figure 5.5 shows the decreasing dielectric constant of the phantom material at 5 GHz. Estimation using a linear trend line shows that the decreasing rate is 1.88 per hour. Phantom material with larger size should have smaller decreasing rate due to smaller surface to volume ratio. Due to the constraint caused by this effect, measurements are conducted within a few minutes. The phantom materials are put back into the container as soon as the measurement is done so that the air exposure effect can be neglected.

Figure 5.5: Phantom material dielectric constant at 5 GHz measured within 12 hours with each measurement conducted every half an hour.

5.3 Phantom Materials in Cylindrical Container

Phantom materials of different content of oil and water are fabricated in polypropylene container of cylindrical shape using Lazebnik's method [39]. The volume of the material in each container is 400 ml. Seven phantom materials are fabricated with 10%, 30%, 50%, 60%, 65%, 70% and 80% of oil. Dielectric measurements are conducted one week after the fabrication when the phantoms have solidified. Lazebnik [39] has shown that the dielectric properties of the breast phantoms are stable from 1 to 7 weeks.

Figure 5.6: Description of dielectric measurement procedure

In order to analyze the dielectric consistency of phantom material, the material is sliced into three layers consisting of four surfaces as shown in Figure 5.6 (surface 2.1 and 2.2 are considered one surface). For each surface, sixteen measurements are taken from four sample groups with each group of measurements taken from a circular region of 1 cm radius. As such, a total sixty four measurements (Acquisition No. 1 - 64) are collected from each phantom.

After measuring each sample group, the probe is cleaned with tissue paper to prevent any oil that may accumulate on it during measurements. After cleaning, measurement of air is taken to ensure that the dielectric constant does not drift away from the baseline. Probe calibration is conducted after every sixteen measurements (one surface) to ensure accuracy. The homogeneity of the phantom material is assessed by comparing the dielectric properties between different regions within the same surface, and between different surfaces within the same phantom. The dielectric constants are plotted in Figure 5.7 with different color for each surface. From bottom surface to top surface, the color representations are red, magenta, blue, and green.

Figure 5.7: Dielectric constants for phantom materials with different concentration of oil (left). Dielectric constant at 5 GHz for measurement from top layer to bottom layer (right).

Figure 5.7 shows that there is large inconsistency in dielectric constant for different surfaces within a phantom material. The surface-to-surface dielectric constants vary greatly for phantom materials with low percentages of oil. This is caused by migration of water over time towards the bottom of the phantom due to gravity before solidification of the materials which occurs 5 days after fabrication. The dielectric constants of the bottom surface tend to be higher due to the concentration of water and the dielectric constants of the top surface tend to be lower due to loss of water.

There is no significant difference for dielectric constants at different regions on the same surface.

5.4 Phantom Materials in Flat Rectangular Container

In order to reduce the dielectric inconsistency of phantom materials, an attempt is made to fabricate materials in another configuration. Phantom materials are fabricated as flat rectangular sheets in order to minimize the effect of water migration and dielectric difference from top to bottom.

The flat container is designed to have dimension of 30 cm height, 20 cm width, and 1 cm thickness as shown in Figure 5.8. The total volume of the flat container is 600 ml. Four sheets with the same percentage (50%) of oil are fabricated to compare the phantom-to-phantom dielectric constants.

Measurements (Acquisition No. 1 - 30) are conducted from bottom to top on a straight line with each measurement taken at every centimeter as shown in Figure 5.8. Measurements are repeated with two lines on the front of the sheet and two lines on the back of the sheet. There are 120 measurements in total.

The measurement data obtained from each sheet is divided into four regions (1-4): region 1 (0-7 cm from bottom), region 2 (7-15 cm from bottom), region 3 (15-23 cm from bottom) and region 4 (23-30 cm from bottom).

Region 1 to 4 are represented by red, magenta, blue, and green color for the dielectric constants plots as shown in Figure 5.9 (left). The plots show that there are no significant differences between different regions from bottom to top of the sheets. The individual measurement line and mean dielectric constants at 5 GHz are plotted as shown in Figure 5.5 (right). The plots show only slight variations in dielectric constant from bottom to top of the sheet.

Figure 5.8: Dielectric sheet in a flat rectangular container

Figure 5.9: Dielectric constants for four phantom material sheets with 50% oil (left). Individual and mean dielectric constants measurements at 5 GHz from the bottom to top of the sheet (right)

5.5 Fabrication of Breast Phantoms

Tissue-mimicking phantom materials are made using Lazebnik's method [39] and results have been discussed in section 5.3 and 5.4.

Six homogeneous breast phantoms are fabricated with different volume (50%, 65%, 80%) of oil. Breast phantoms are fabricated in cylindrical polypropylene containers of 600 ml. The containers dimension is 8 cm height and 10 cm diameter. The dielectric constants estimated from Lazebnik [39] are shown in Table 5.4. Dielectric measurement results will be discussed in section 5.6.

Table 5.4: Homogeneous breast phantoms fabricated with different percentage of oil

Breast Phantom	Volume % of oil in phantom	Dielectric constant at 5 GHz
Homo-80	80%	8
Homo-65	65%	16
Homo-50	50%	24

Seven heterogeneous breast phantoms are fabricated by mixing phantom materials made of different volume of oil. Clutters are made of material with low percentage of oil to simulate the glandular tissues, whereas matrix to hold the clutters is made of material with 80% of oil to simulate the adipose tissues in human breast. Clutters are prepared by chopping the high dielectric phantom materials to size smaller than 5mm. The compositions of different breast phantoms are shown in Table 5.5.

The fabrication of heterogeneous breast phantom is illustrated in Figure 5.10. A thin layer of matrix material (80% oil) is poured into the phantom container. Then a thin layer of clutter material (50% oil) is deposited on the thin layer of matrix material. Another thin layer of matrix material is poured onto the thin layer of clutter material. The process is repeated until the phantom materials fully filled the container.

Figure 5.10a: Pouring the low dielectric matrix material which is to simulate the adipose tissue in human breast.

Figure 5.10b: Pouring the high dielectric clutter material which is to simulate the glandular and fibroconnective tissues in human breast.

Phantoms Hetero-17, Hetero-25, Hetero-33, and Hetero-50 are fabricated by fixing the clutters dielectric constant and varying the percentage of clutters to simulate the uncertainty in volume of glandular and fibroconnective tissues. Phantoms Hetero-70, Hetero-65, and Hetero-60 are fabricated by fixing percentage of clutters and varying the clutters dielectric constant to simulate the uncertainty in averaged breast dielectric properties. The compositions of different breast phantoms are shown in Table 5.5. The dielectric constants are estimated from Lazebnik [39]. Dielectric properties measured using the Agilent probe are discussed in section 5.7.

Table 5.5: Compositions of seven fabricated heterogeneous breast phantoms

Phantom	Volume % of oil in clutters	Volume % of clutters in phantom	Volume % of oil in phantom	Dielectric constant at 5 GHz
Hetero-17	50%	17%	75%	10
Hetero-25	50%	25%	73%	11
Hetero-33	50%	33%	70%	13
Hetero-50	50%	50%	65%	16
Hetero-70	70%	50%	75%	10
Hetero-65	65%	50%	73%	11
Hetero-60	60%	50%	70%	13

5.6 Dielectric Properties of Homogeneous Breast Phantoms

Six homogeneous breast phantoms are fabricated with different percentages (50%, 65%, 80%) of oil. Unlike the fabrication of phantom materials in section 5.3, breast phantom is fabricated with volume of 600 ml to fully occupy the polypropylene container. The breast phantom is turned top-to-bottom every half an hour for the next 6 hours after fabrication. The procedure is to avoid the water migration towards the bottom of the phantom.

Similar to the procedure in section 5.3, a total of 128 dielectric measurements are taken with the phantoms sliced into 7 layers with 8 surfaces.

Analysis of variance (ANOVA) is conducted to analyze the layer-to-layer difference. ANOVA result of 1 indicates no difference between layers, whereas 0 indicates distinct difference between layers. Results in Figures 5.11 to 5.16 show that turning the phantoms top-to-bottom and bottom-to-top does not resolve the water migration problem. However if the top and bottom layers are excluded from analysis, the layer-to-layer difference is not significant.

When considering the six layers in the middle part of phantoms, ANOVA results are between 0.005 and 0.91, whereas when considering all layers, ANOVA results are smaller than 0.001. For UWB imaging, the top and bottom layers should not affect the performance since the Thales directional antenna is used. In section 5.8, we shall examine the variability by looking at the absolute variance rather than the relative variance of dielectric constant.

In this section, all the analysis is conducted only for dielectric constant at 5 GHz. This is because the variability of dielectric constants is linear across the whole frequency of measurement, as can be seen in section 5.3 and 5.4. Histograms of dielectric constants at 5 GHz are plotted at right side of Figure 5.11 to 5.15. Results show that the dielectric constants are normally distributed.

Figure 5.11a: Breast phantom Homo-50a. Box plot (left) for 8 surfaces measurements. Histogram (right) of 128 measurements

Figure 5.11b: Breast phantom Homo-50a. Box plot (left) excluding the top and bottom layers. Histogram (right) of 96 measurements from middle layers

Figure 5.12a: Breast phantom Homo-50b. Box plot (left) for 8 surfaces measurements. Histogram (right) of 128 measurements

Figure 5.12b: Breast phantom Homo-50b. Box plot (left) excluding the top and bottom layers. Histogram (right) of 96 measurements from middle layers

Figure 5.13a: Breast phantom Homo-65a. Box plot (left) for 8 surface measurements. Histogram (right) of 128 measurements

Figure 5.13b: Breast phantom Homo-65a. Box plot (left) excluding the top and bottom layers. Histogram (right) of 96 measurements from middle layers

Figure 5.14a: Breast phantom Homo-65b. Box plot (left) for 8 surfaces measurements. Histogram (right) of 128 measurements

Figure 5.14b: Breast phantom Homo-65b. Box plot (left) excluding the top and bottom layers. Histogram (right) of 96 measurements from middle layers

Figure 5.15a: Breast phantom Homo-80a. Box plot (left) for 8 surfaces measurements. Histogram (right) of 128 measurements

Figure 5.15b: Breast phantom Homo-80a. Box plot (left) excluding the top and bottom layers. Histogram (right) of 96 measurements from middle layers

Figure 5.16a: Breast phantom Homo-80b. Box plot (left) for 8 surfaces measurements. Histogram (right) of 128 measurements

Figure 5.16b: Breast phantom Homo-80b. Box plot (left) excluding the top and bottom layers. Histogram (right) of 96 measurements from middle layers

5.7 Dielectric Properties of Heterogeneous Breast Phantoms

Seven heterogeneous breast phantoms are fabricated as discussed in Section 5.5. Four phantoms, namely Hetero-17, Hetero-25, Hetero-33, and Hetero-50, are fabricated by fixing the clutters dielectric constant and varying the percentage of clutters. Three phantoms, namely Hetero-70, Hetero-65, and Hetero-60, are fabricated by fixing the percentage of clutters and varying the clutters dielectric constant. Dielectric measurements are conducted with the same procedure as in section 5.6.

Measurement results for dielectric constant of 5 GHz are shown with box plots and histograms in Figure 5.17 to 5.23. The distributions of the dielectric constants are skewed to the left. The distribution skewness is higher for breast phantom with lower volume percentage of clutters of the same dielectric constant. This is because the dielectric measurements are conducted with random selection of samples. In most of the instances, the probe is measuring the matrix, and in few instances the probe is measuring the clutter.

The distributions show that the variability of the dielectric properties in breast tissues can be successfully simulated with clutters of single dielectric constant. By chopping the phantom material to small size ($<5\text{mm}$), the distinct dielectric constant of small clutters will be averaged with the surrounding low dielectric matrix medium. This is because the small clutters are considered microscopic at microwave frequencies. Thus a smooth distribution from low to high dielectric constant can be created to represent the breast dielectric variability.

The dielectric constants of the top and bottom layers are low because the clutters are totally covered by the matrix material. During the fabrication process, a thin layer of matrix material is first poured at the bottom of the phantom, and finally the top of the phantom is covered with a thin layer of matrix material.

Figure 5.17: Breast phantom Hetero-17. Box plot (left) for 8 surfaces measurements. Histogram (right) of 128 measurements

Figure 5.18: Breast phantom Hetero-25. Box plot (left) for 8 surfaces measurements. Histogram (right) of 128 measurements

Figure 5.19: Breast phantom Hetero-33. Box plot (left) for 8 surfaces measurements. Histogram (right) of 128 measurements

Figure 5.20: Breast phantom Hetero-50. Box plot (left) for 8 surfaces measurements. Histogram (right) of 128 measurements

Figure 5.21: Breast phantom Hetero-60. Box plot (left) for 8 surfaces measurements. Histogram (right) of 128 measurements

Figure 5.22: Breast phantom Hetero-65. Box plot (left) for 8 surfaces measurements. Histogram (right) of 128 measurements

Figure 5.23: Breast phantom Hetero-70. Box plot (left) for 8 surfaces measurements. Histogram (right) of 128 measurements

5.8 Analysis of Dielectric Variability

In this section, analysis of dielectric variability is conducted by comparing the variance of different measurement groups within a phantom.

For cylindrical phantom materials, the variances are calculated for 4 samples from the same circular region of 1 cm radius, 16 samples for each surface, and a total of 64 samples for each phantom. For flat rectangular phantom materials, the variances are calculated for 30 samples along each measurement line, 30 samples for each rectangular region, and a total of 120 samples for each phantom.

Results in Table 5.6 show that variances for each phantom are large for phantom materials with lower concentration of oil. The results are expected due to the large layer-to-layer differences as discussed in section 5.6. For variances of 1 cm radius circular region and the same surface, there is not much difference showing that the dielectric constants for each layer are relatively consistent.

For sheet materials fabricated with 50% oil, results in Table 5.7 show that variances for each phantom are small. That is the dielectric constants are consistent throughout the sheet. The sheet-to-sheet differences as shown in Table 5.8 are small showing that the fabrication procedure is able to produce consistent phantom materials.

Table 5.6: Variances of different groups' measurements for phantom materials

Material	1 cm radius circular region	Surface	Phantom	Mean
10 % oil	1.24	1.43	269.5	40.58
30 % oil	1.01	1.58	183.5	33.08
50 % oil	1.40	3.54	41.43	22.52
60 % oil	0.50	0.80	6.01	14.63
65 % oil	0.55	1.34	2.99	12.88
70 % oil	0.51	0.86	1.44	11.31
80 % oil	0.08	0.87	1.11	7.72

Table 5.7: Variances of different groups' measurements for sheets with 50% oil

Material	Measurement Line	Rectangular Region	Phantom
Sheet 1	2.0	1.7	1.9
Sheet 2	1.1	1.2	1.4
Sheet 3	3.4	2.9	3.2
Sheet 4	0.9	0.8	0.9

Table 5.8: Means and deviations from overall mean (16.36) for sheet 1 to sheet 4

Material	Mean	Deviation from Overall Mean	Variance
Sheet 1	15.53	0.83	1.18
Sheet 2	15.35	1.01	1.18
Sheet 3	17.01	0.65	1.18
Sheet 4	17.55	1.19	1.18

For homogeneous breast phantoms, the analysis is conducted similar to cylindrical phantom materials except that there are more measurements for each phantom. The variances are calculated for 4 samples from the same circular region of 1 cm radius, 16 samples for each surface, and a total of 128 samples for each phantom.

The calculated mean and variance for each measurement group are shown in Figure 5.24. The values shown represent the mean $\pm$ variance. Eight boxes represent measurements of eight surfaces. The values at the center of the box are the mean and variance for the surface. 32 variances from circular regions of 1 cm radius and 8 variances of surfaces measurements are averaged and the results are summarized in Table 5.9.

For heterogeneous breast phantoms, similar analysis is conducted and the results are shown in Table 5.10.

Figure 5.24: Mean and variance for each group of measurements for a homogeneous phantom. Each text box represents each surface measurements.

Figure 5.25: Box plot for each surface measurements for a homogeneous phantom.

Table 5.9: Variances for homogeneous breast phantoms

Phantom	1 cm radius circular region	Surface	Phantom	Mean
Homo-50a	3.90	6.22	14.80	18.54
Homo-50b	2.76	3.22	24.58	21.44
Homo-65a	0.69	0.99	2.19	12.20
Homo-65b	0.46	0.81	1.45	13.42
Homo-80a	0.32	0.64	0.75	7.44
Homo-80b	0.28	0.33	0.41	8.93
Top and bottom layers excluded				
Homo-50a	4.70	7.26	7.61	17.32
Homo-50b	3.47	3.13	3.01	21.27
Homo-65a	0.86	0.91	1.03	11.69
Homo-65b	0.54	0.86	0.87	12.99
Homo-80a	0.33	0.55	0.62	7.42
Homo-80b	0.34	0.40	0.45	9.06

Table 5.10: Variances for heterogeneous breast phantoms

Phantom	1 cm radius circular region	Surface	Phantom	Mean
Hetero-17	7.56	7.68	8.28	9.86
Hetero-25	8.60	8.98	8.86	11.32
Hetero-33	11.46	10.95	11.98	11.85
Hetero-50	17.69	16.85	19.35	13.23
Hetero-60	6.78	6.25	7.42	11.28
Hetero-65	3.32	3.48	3.80	10.45
Hetero-70	1.77	1.81	2.22	9.97

Results in Table 5.9 show that there is improvement for the layer-to-layer consistency with the fabrication procedure used in homogeneous breast phantom as compared to phantom materials as shown in Table 5.6. By excluding the top and bottom layers of the breast phantoms, the phantom variances are significantly smaller, showing that the middle layers dielectric constants are consistent.

Table 5.10 shows the variance of different measurement groups for heterogeneous breast phantoms. There is no significant difference for variances of 1 cm radius circular region, surface, and phantom showing that the clutters are well distributed during fabrication.

Variability of the breast phantoms fabricated instead of variance provides a better indicator of the dielectric properties of the phantoms. The variability is calculated as two times the standard deviation divided by the mean and the results are shown in Table 5.11 and 5.12. The results show that the fabricated heterogeneous breast phantoms provide the required variability of human breast for subsequent UWB imaging.

The dielectric variability for homogeneous breast phantom is below 30% and above 30% for heterogeneous breast phantom. There is slight overlapping in variability percentage for phantoms Homo-80 and Hetero-70 due to imperfect fabrication control.

Table 5.11: Variability of homogeneous breast phantoms

Phantom	1 cm radius circular region	Surface	Phantom	Mean
Homo-50a	25.0%	31.1%	31.9%	18.54
Homo-50b	17.5%	16.6%	16.3%	21.44
Homo-65a	15.9%	16.3%	17.4%	12.20
Homo-65b	11.3%	14.3%	14.4%	13.42
Homo-80a	15.5%	20.0%	21.2%	7.44
Homo-80b	12.9%	14.0%	14.8%	8.93

Table 5.12: Variability of heterogeneous breast phantoms

Phantom	1 cm radius circular region	Surface	Phantom	Mean
Hetero-17	55.8%	56.2%	58.4%	9.86
Hetero-25	51.8%	52.9%	52.6%	11.32
Hetero-33	57.1%	55.8%	58.4%	11.85
Hetero-50	63.6%	62.1%	66.5%	13.23
Hetero-60	46.2%	44.3%	48.3%	11.28
Hetero-65	34.9%	35.7%	37.3%	10.45
Hetero-70	26.7%	27.0%	29.9%	9.97

Chapter 6:

UWB Imaging on Realistic Breast Phantoms

In this chapter, experiments of UWB imaging on realistic breast phantoms are presented. Breast phantoms are realistic and comparable to human breast in term of dielectric properties. The experiments are designed to incorporate the recent large scale dielectric properties measurements of normal and cancer breast tissues by Lazebnik [48, 49].

Unlike commonly assumed dielectric properties in most simulation studies, the new findings show that the breast dielectric constant and variability are much higher. Previous assumption set the breast dielectric constant as 9 and variability as 10%. The assumption is based on the extrapolation of small scale measurement studies given in Chapter 2 to frequency range above 3GHz using the Debye model [11].

Section 6.1 describes the experimental design based on Lazebnik's [48, 49] breast dielectric properties. The UWB imaging experiments on homogeneous and heterogeneous breast phantoms are given in section 6.2 and section 6.3 respectively. Section 6.4 discusses the limiting factors for tumor detection. Section 6.5 presents tumor response analysis to study the signal loss caused by different dielectric properties. Finally, estimation of the detection resolution with different limiting factors is given in section 6.6.

6.1 Experimental Design

Three homogeneous and seven heterogeneous breast phantoms are fabricated. The fabrication of breast phantoms with different volume percentage of clutters and different clutters dielectric constants is discussed in Chapter 5. The experiment is designed so that it is significant in two ways:

1. UWB imaging is conducted using homogeneous breast phantoms with similar dielectric constants of human breast compared to other reported experiments which used much lower dielectric constants.
2. UWB imaging is conducted using heterogeneous breast phantoms which simulate the heterogeneous nature of human breast which has not been reported thus far.

The breast phantoms with their mean dielectric constants at 5 GHz and variability are shown in Table 6.1. All heterogeneous breast phantoms are fabricated using phantom material with dielectric constant of 8 as the medium matrix. The range is calculated from the difference between clutters dielectric constant and the matrix dielectric constant. The variability is calculated with range divided by two times mean dielectric constant. Note that the variability may not be centered on the mean dielectric constant since the breast phantoms with less percentage of clutters will have dielectric constant distribution skewed to the left as shown in Chapter 5.

Table 6.1: Dielectric constants at 5 GHz and variability for fabricated homogenous and heterogeneous breast phantoms

Phantom	Mean dielectric constant	Dielectric constant	Range of dielectric constant	Variability
Homo-80	8	8	0	0%
Homo-65	16	16	0	0%
Homo-50	24	24	0	0%
Hetero-17	10	8 – 24	16	±80%
Hetero-25	11	8 – 24	16	±73%
Hetero-33	13	8 – 24	16	±62%
Hetero-50	16	8 – 24	16	±50%
Hetero-60	13	8 – 20	12	±46%
Hetero-65	11	8 – 16	8	±36%
Hetero-70	10	8 – 12	4	±20%

In all the reported experimental studies, only dielectric contrast between breast and tumor is simulated. The breast phantoms are made of materials with much lower dielectric constants and much lower dielectric loss than human breasts, as shown in Table 2.4 in Chapter 2. It is assumed that the imaging experiments on human breast can be simulated by sending low power signals to simulate the loss effects. However, for lower dielectric constant materials, the estimated speed is less sensitive to estimation error since the signals are delayed by smaller time shifts. Dispersion effects are also less significant with lower dielectric constant.

In addition, for lower dielectric loss materials, the tumor response to incident pulse amplitude ratio is of the order of 10^2 , whereas for human breast like dielectric loss material the ratio is of the order of 10^3 . This poses a challenging problem for early time and skin artifact removal because of the much higher SNR required.

In heterogeneous breast phantoms, signals strengths are reduced when the tumor response travels through the medium and is scattered by the clutters. Apart from that, the strong clutter responses become dominant in the breast image and make the weak tumor response undetectable.

Figure 6.1 and 6.2 summarize the experiments done in this chapter as compared to published results. From Lazebnik's [48, 49] large scale studies, the median breast tissue dielectric constant is 34 and the range of dielectric constant is 44 with variability of $\pm 65\%$. The variability is much higher than the commonly assumed value of $\pm 10\%$ in simulation studies.

Figure 6.1: Dielectric constant of breast tissues

The commonly assumed tumor dielectric constant is 50 which is close to 49 given by Lazebnik [48, 49]. In our experiment design, the breast phantom dielectric constants are in the range 8 to 24 which is included in the range 4 to 48 reported by Lazebnik [48, 49]. This is because above 24, signal loss is very high and beyond the detection capability of the experimental setup.

Figure 6.2: Variability of breast tissues dielectric constant

Four materials are used as the tumor stimulant as shown in Table 6.2. The tumor is inserted into the breast phantoms at depth of 4 cm and 2 cm off center and attached to a polyester thread. The time required for each imaging procedure is four hours with two hours for scanning the tumor-bearing phantom and two hours for scanning the tumor-free phantom.

Table 6.2: Dimensions and dielectric constants of materials used as tumor simulant

Tumor	Diameter (mm)	Length (mm)	Dielectric constant
Metal rod	2.5	80	Perfect reflector
Metal pin	1.0	8	Perfect reflector
Phantom material	2.0	10	50
Phantom material	4.0	10	50

Figure 6.3: Materials used as tumor stimulants in breast phantom

Metal rod is used to simulate the breast tumor in 2D scenario, which is commonly assumed by Finite Difference Time Domain (FDTD) studies [15, 20, 30], to calibrate the experimental setup. Metals are used because it reflects all signals and the signal strength is above noise floor.

For each imaging experiment, subtraction and averaging methods are applied before image formation. Subtraction method is applied by subtracting the data from tumor-bearing phantom with data from tumor-free phantom. Subtraction method removes everything including the clutters responses from the phantom, thus the remaining signals are clean tumor responses. The method is not applicable in practice but only for benchmarking purpose to compare with the averaging method.

6.2 Imaging of Homogeneous Breast Phantoms

Imaging results for three homogeneous breast phantoms, namely Homo-50, Homo-65, and Homo-80, are shown in Figure 6.4, 6.5, and 6.6. The names of the phantoms indicate the percentage of oil used in fabrication. Images formed with subtraction method are the left figures to be compared to the right figures for averaging method. The imaging results are summarized in Table 6.3.

Table 6.3: Detectability of tumor in different homogeneous breast phantoms

Phantom	Method	Metal rod	Metal pin	2mm tumor	4mm tumor
Homo-50	Subtraction	Yes	No	No	Yes
	Averaging	Yes	No	No	No
Homo-65	Subtraction	Yes	Yes	No	Yes
	Averaging	Yes	No	No	No
Homo-80	Subtraction	Yes	Yes	Yes	Yes
	Averaging	Yes	No	No	Yes

Figure 6.4: Imaging results for breast phantom Homo-50

Figure 6.5: Imaging results for breast phantom Homo-65

Figure 6.6: Imaging results for breast phantom Homo-80

6.3 Imaging of Heterogeneous Breast Phantoms

Imaging results for seven heterogeneous breast phantoms, namely Hetero-17, Hetero-25, Hetero-33, Hetero-50, Hetero-60, Hetero-65 and Hetero-70, are shown in Figure 6.7 to Figure 6.13. The names for 17, 25, 33, and 50 indicate the percentage of clutters used in fabrication, with clutters made of 50% of oil. The names for 60, 65, and 70 indicate the percentage of oil in clutters used in fabrication, with 50% of clutters. Image formed with subtraction method are the left figures to be compared to the right figures formed by averaging method. . The imaging results are summarized in Table 6.4.

Table 6.4: Detectability of tumor in different heterogeneous breast phantoms

Phantom	Method	Metal rod	Metal pin	2mm tumor	4mm tumor
Hetero-17	Subtraction	Yes	Yes	Yes	Yes
	Averaging	Yes	No	No	Yes
Hetero-25	Subtraction	Yes	Yes	Yes	Yes
	Averaging	Yes	No	No	Yes
Hetero-33	Subtraction	Yes	Yes	Yes	Yes
	Averaging	Yes	No	No	No
Hetero-50	Subtraction	Yes	Yes	Yes	Yes
	Averaging	Yes	No	No	No
Hetero-60	Subtraction	Yes	Yes	Yes	Yes
	Averaging	Yes	No	Yes	Yes
Hetero-65	Subtraction	Yes	Yes	Yes	Yes
	Averaging	Yes	No	No	Yes
Hetero-70	Subtraction	Yes	Yes	Yes	Yes
	Averaging	Yes	No	No	Yes

Figure 6.7: Imaging results for breast phantom Hetero-17

Figure 6.8: Imaging results for breast phantom Hetero-25

Figure 6.9: Imaging results for breast phantom Hetero-33

Figure 6.10: Imaging results for breast phantom Hetero-50

Figure 6.11: Imaging results for breast phantom Hetero-60

Figure 6.12: Imaging results for breast phantom Hetero-65

Figure 6.13: Imaging results for breast phantom Hetero-70

6.4 Limiting Factors for Tumor Detectability

There are factors limiting the detectability of tumor, namely system, algorithm and breast phantom limitations. In order to study the effects of different limitations, signal strength of tumor response and artifact strength caused by different limitations are to be analyzed separately. To isolate the signals from the effects of these limitations, subtraction method is used since it is able to remove everything to give clean tumor responses. Whereas to investigate the artifact strength due to different limitations, data sets obtained from tumor-free phantoms are used.

The analysis helps to identify which factor is the major limiting factor for breast tumor detection. And based on these limitations, we are able to estimate the detection resolution of breast tumor. The artifact amplitudes for different limiting factors are shown in Table 6.5 to 6.7. Pixel intensity is the variance of windowed summed signal.

System limitation is caused by the limited dynamic range of the oscilloscope. It is calculated from the artifact amplitudes in homogeneous tumor-free breast phantom images formed with subtraction method.

Table 6.5: Artifact amplitude due to system limitation

Breast Phantom	Artifact Pixel Intensity	Summed Signal Amplitude	Max Signal Amplitude
Homo80	0.00006	0.027	0.00035
Homo65	0.00003	0.020	0.00029
Homo50	0.00023	0.051	0.00043
Average	0.00011	0.033	0.00036

Algorithm limitation is due the simple averaging processing used. Artifacts present in the image are due to slight unsymmetrical placement or shape of the phantom. It is calculated from the artifact amplitudes in homogeneous tumor-free breast phantom images formed with averaging method.

Table 6.6: Averaging artifacts due to algorithm limitation

Breast Phantom	Artifact Pixel Intensity	Summed Signal Amplitude	Max Signal Amplitude
Homo80	0.0074	0.30	0.0017
Homo65	0.0058	0.28	0.0022
Homo50	0.0025	0.22	0.0018
Average	0.0052	0.27	0.0019

Breast phantom limitation is due the presence of clutters in heterogeneous breast phantoms. It is calculated from the clutter pixels intensity in heterogeneous tumor-free breast phantom images formed with averaging method.

Table 6.7: Clutters responses due to breast phantom limitation

Breast Phantom	Clutter Pixel Intensity	Summed Signal Amplitude	Max Signal Amplitude
Hetero17	0.034	0.60	0.0019
Hetero25	0.025	0.56	0.0024
Hetero33	0.046	0.72	0.0029
Hetero50	0.008	0.30	0.0022
Hetero-60	0.004	0.23	0.0013
Hetero-65	0.010	0.35	0.0018
Hetero-70	0.003	0.21	0.0012
Average	0.019	0.42	0.0020

6.5 Tumor Response Analysis

When tumor pixel intensity is higher than the artifact intensity, tumor will be successfully detected. If the tumor pixel intensity is lower than the artifact intensity, tumor is undetectable.

For homogeneous breast phantoms with higher dielectric constants, decrease in signal strength is caused by decrease in dielectric contrast between tumor and medium. Another factor is higher loss in the medium. For heterogeneous breast phantoms, decrease in signal strength is caused by scattering of tumor responses by clutters when travelling through the medium.

Steel rod and gold-plated pin are used to simulate the tumor because metal reflects all signals and the signal strength is above noise floor. For tumor simulated using high dielectric phantom material, tumor responses vary largely due to deformation of shape and changing of dielectric constant during experiment.

Ratios of summed signal amplitudes over max signal amplitudes for different tumor simulants are given in Table 6.8. The ratio is an indicator of how good the signals are aligned. If all the individual signals are of equal amplitude, the ratio would be 360 because of the 360 antennas used. Due to loss in the medium, the best ratios are ranging from 100 to 117. Ratio below 50 indicates poor alignment of signals or the signals are below noise floor.

Table 6.8: Ratios of summed signal amplitudes over max signal amplitudes for different tumor simulants

Breast Phantom	Metal Rod	Metal Pin	2 mm Tumor	4 mm Tumor
Homo-80	117	76	121	113
Homo-65	90	75	86	83
Homo-50	62	38	17	63
Hetero-17	117	105	54	105
Hetero-25	106	76	92	102
Hetero-33	103	77	117	100
Hetero-50	89	58	75	84
Hetero-60	92	54	72	80
Hetero-65	89	50	62	82
Hetero-70	108	64	70	102

Pixel intensities for the four tumor simulants are plotted versus different parameters of the fabricated breast phantoms. As shown in Figure 6.14, higher dielectric constant is the major cause of tumor signal strength reduction. For breast phantom with averaged dielectric constant higher than 24, detection of small tumor is almost impossible since the system noise floor is 0.0001 as discussed in section 6.4.

Reduction in tumor signal strength caused by presence of clutters is less severe as shown in Figure 6.15. For the plot of pixel intensity versus dielectric variability as shown in Figure 6.16, the tumor signal strength reduction is not significant. This is because the effects of the other parameters, such as averaged dielectric constant and percentage of clutters, of the breast phantoms cannot be isolated in the analysis.

Figure 6.14: Pixel intensity versus dielectric contrast for homogeneous breast phantoms

Figure 6.15: Pixel intensity versus percentage of clutters for heterogeneous breast phantoms

Figure 6.16: Pixel intensity versus dielectric variability for heterogeneous breast phantoms

6.6 Estimation of Detection Resolution

Detection resolution is the size of the smallest detectable tumor which shows up as the brightest pixel in the breast phantom. The term differs from image resolution which is defined as the smallest separation distance between two tumors where the tumors are still distinguishable in the image.

Detection resolution is estimated based on the results obtained with four different tumor simulants as discussed in this chapter. Results obtained with metal tumor stimulant are used to estimate the tumor signal strength of tumor material of the same size with dielectric constant of 50. Estimation is made based on the reflection coefficient of tumor in breast medium, which is 0.4. Pixel intensity is estimated by multiplying factor of 0.16. For tumors of different size in phantoms of different dielectric constants, the tumor signal strength is estimated by interpolation of the available data. Results are shown in Table 6.9 to 6.12 for three different limiting factors.

Signal-to-noise ratios (SNR) as shown in Table 6.9 are calculated with artifact pixel intensity of 0.0001 as estimated in section 6.4. Results show that system limitation is not the major limiting factor for detection of breast cancer. The experimental setup dynamic range is 72 dB (12 bits) with noise level of 0.3 mV as discussed in Chapter 4. Detection of 2 mm tumor is possible with breast phantom with dielectric constant up to 20.

Table 6.9: SNR for tumors of different sizes due to system limitation

Dielectric constant	8	12	16	20	24
2D scenario					
2mm tumor	30.3 dB	21.0 dB	15.7 dB	13.8 dB	11.2 dB
3D scenario					
5mm tumor	27.0 dB	23.5 dB	18.3 dB	13.6 dB	6.6 dB
4mm tumor	23.7 dB	20.3 dB	15.0 dB	10.4 dB	4.4 dB
3mm tumor	21.8 dB	17.1 dB	13.0 dB	8.0 dB	2.6 dB
2mm tumor	18.5 dB	14.4 dB	9.6 dB	5.1 dB	-0.4 dB
1mm tumor	8.9 dB	5.3 dB	2.6 dB	-0.4 dB	-3.4 dB

Algorithm limitation is due to averaging artifacts. Signal-to-noise ratios (SNR) as shown in Table 6.10 are calculated with artifact pixel intensity of 0.005. For an ideal breast phantom scenario which works perfectly for averaging algorithm, the resultant noise level is 2mV as discussed in section 6.4. Detection of 2 mm tumor stimulant is possible with breast phantom with dielectric constant up to 8.

Table 6.10: SNR for tumors of different sizes due to algorithm limitation

Dielectric constant	8	12	16	20	24
2D scenario					
2mm tumor	13.5 dB	4.3 dB	1.0 dB	-2.9 dB	-5.6 dB
3D scenario					
5mm tumor	10.3 dB	6.7 dB	1.6 dB	-3.2 dB	-10.2 dB
4mm tumor	6.9 dB	3.6 dB	-1.7 dB	-6.4 dB	-12.4 dB
3mm tumor	5.0 dB	0.4 dB	-3.7 dB	-8.7 dB	-14.1 dB
2mm tumor	1.7 dB	-2.4 dB	-7.2 dB	-11.6 dB	-17.2 dB
1mm tumor	-7.9 dB	-11.4 dB	-14.1 dB	-17.2 dB	-20.2 dB

Breast phantom limitation is due to the presence of clutters. The signal-to-clutter ratios (SCR) as shown in Table 6.11 and 6.12 are calculated with clutter pixel intensity as estimated in section 6.4. Detection of 4mm tumor is possible in breast phantom with percentage of clutters up to 25% with clutters dielectric constant of 24, or in breast phantom with clutters dielectric constant up to 20, with percentage of clutters of 50%.

Table 6.11: Breast limited SCR for tumors of different sizes in phantoms with 50% clutters of different dielectric constants

Dielectric of clutters	12	16	20	24
2D scenario				
2mm tumor	11.8 dB	8.7 dB	4.2 dB	-2.5 dB
3D scenario				
5mm tumor	10.3 dB	7.4 dB	1.9 dB	-3.3 dB
4mm tumor	8.9 dB	5.6 dB	0.9 dB	-6.9 dB
3mm tumor	5.3 dB	1.5 dB	-3.4 dB	-11.7 dB
2mm tumor	0.4 dB	-3.7 dB	-8.5 dB	-16.4 dB
1mm tumor	-6.6 dB	-9.5 dB	-14.3 dB	-21.7 dB

Table 6.12: Breast limited SCR for tumors of different sizes in phantoms with different percentage of clutters of the same dielectric constants of 24

Percentage of clutters	17%	25%	33%	50%
2D scenario				
2mm tumor	9.1 dB	2.8 dB	1.7 dB	-2.5 dB
3D scenario				
5mm tumor	8.7 dB	4.0 dB	0.3 dB	-3.3 dB
4mm tumor	5.7 dB	0.3 dB	-3.6 dB	-6.9 dB
3mm tumor	0.5 dB	-4.4 dB	-8.5 dB	-11.7 dB
2mm tumor	-5.2 dB	-10.0 dB	-13.4 dB	-16.4 dB
1mm tumor	-12.6 dB	-16.3 dB	-19.2 dB	-21.7 dB

Chapter 7: Conclusion and Future Work

Experiments on pulse-based UWB microwave breast imaging with realistic tissue mimicking breast phantoms are conducted. Experimental issues which are important for the development of pulse-based UWB breast cancer detection prototype and the compensation methods are studied. These includes pulse instability, loss compensation, narrowband frequencies interference, limited dynamic range, phantom positioning error, averaging number and antenna number. Detection capability is significantly improved by the compensation methods.

The breast phantoms used in the experiments are realistic and comparable to human breast in term of dielectric properties. Unlike previously reported experiments that use lower (compared to human breast tissue) dielectric homogeneous materials as the breast medium, the experiments are significantly different in two ways. Firstly, dielectric loss and dielectric constant of the breast phantoms used in the experiments are close to human breasts. Secondly, heterogeneous breast phantoms are fabricated by mixing phantom materials of different dielectric constants to simulate the glandular tissue clutters in adipose tissue matrix of the human breasts.

More realistic experiments are designed based on the new results from large scale dielectric measurements on normal and malignant breast tissues by Lazebnik [48, 49]. The breast tissue dielectric constant is ranging from 4 to 48 with dielectric variability of 65% which is much larger than previously assumed 10% variability.

Extensive experiments have been conducted with ten breast phantoms to simulate the wide variability in breast dielectric properties. Three homogeneous breast phantoms with different averaged dielectric constants and seven heterogeneous breast phantoms with different percentages of clutters are fabricated.

Experiment results show that the experimental setup is capable of detecting 2 mm tumor in phantoms with averaged dielectric constant of 20. The major limiting factor for tumor detection in homogeneous breast phantoms with higher averaged dielectric constant is signal loss.

For heterogeneous breast phantom with higher percentage of clutters, the major limiting factor for tumor detection is dominant clutters responses in the breast image. Detection of 4 mm tumor is possible in phantoms with 50% clutters of lower dielectric constant, or in phantoms with 25% clutters of higher dielectric constant.

In future study, advanced signal processing techniques and algorithms such as Microwave in Space Time Beamforming (MIST), Robust Capon Beamforming (RCB), and skin artifacts removal algorithms can be implemented to reduce the noise and clutters responses. These algorithms may further improve the detectability of breast tumor in heterogeneous breast medium.

More realistic breast phantoms can be fabricated with physiological structures such as glandular tissues, skin, and nipples to better represent the human breast. These can be achieved using phantom materials of different dielectric properties chopped into long or stringy pieces to simulate different structures in the breast. Successful detection of breast tumor in physiologically realistic breast phantom will move the research in UWB for breast cancer detection to clinical trial level.

References

- [1] FCC (GPO) Title 47, Section 15 of the Code of Federal Regulations SubPart F: Ultra-wideband, Aug. 2003
- [2] D. J. Daniels, "Surface-penetrating radar," *Electronics & Communication Engineering Journal*, vol.8, no.4, pp.165-182, Aug. 1996
- [3] J. Ahmedin, S. Rebecca, W. Elizabeth, H. Yongping, X. Jiaquan, M. Taylor, and T. J. Michael, "Cancer Statistics 2008," *CA Cancer Journal for Clinicians*, 58: 71-96, 2008.
- [4] F. P. Michael, P. Ruth, C. A. Kathleen, L. Chenxiong, and F. J. Eric, "Age-conditional probabilities of developing cancer," *Statistics in Medicine*, 22(11): 1837-1848, 2003.
- [5] American Cancer Society, Breast Cancer Survival Rates by Stage, [Online]. Available: www.cancer.org.
- [6] National Cancer Institute, Surveillance, Epidemiology, and End Results (SEER) Program, Fast Stats: Breast Cancer: Survival & Stage, [Online]. Available: www.seer.cancer.gov.
- [7] J. T. Bushberg, *The Essential Physics of Medical Imaging*, Philadelphia: Lippincott Williams & Wilkins, 2002.
- [8] J. G. Elmore, M. B. Barton, V. M. Mocerri, S. Polk, P. J. Arena, and S. W. Fletcher, "Ten-year risk of false positive screening mammograms and clinical breast examinations," *New England Journal of Medicine*, vol. 338, no. 16, pp. 1089–1096, 1998.
- [9] P. T. Huynh, A. M. Jarolimek, and S. Daye, "The false-negative mammogram," *Radiographics*, vol. 18, no. 5, pp. 1137–1154, 1998.
- [10] S. J. Nass, I. C. Henderson, and J. C. Lashof, *Mammography and Beyond: Developing Technologies for the Early Detection of Breast Cancer*, Committee on Technologies for the Early Detection of Breast Cancer, Washington, D.C.: Institute of Medicine, National Academy Press, 2000.
- [11] S. C. Hagness, A. Taflove, and J. E. Bridges, "Two-Dimensional FDTD Analysis of a Pulsed Microwave Confocal System for Breast Cancer Detection: Fixed-Focus and Antenna-Array Sensors," *IEEE Transactions on Biomedical Engineering*, vol. 45, no. 12, pp. 1470-1479, Dec. 1998.
- [12] S. C. Hagness, A. Taflove, and J. E. Bridges, "Three-Dimensional FDTD Analysis of a Pulsed Microwave Confocal System for Breast Cancer Detection: Design of an Antenna-Array Element," *IEEE Transactions on Antennas and Propagation*, vol. 47, no. 5, pp. 783-791, May 1999.
- [13] X. Li and S. C. Hagness, "A Confocal Microwave Imaging Algorithm for Breast Cancer Detection," *IEEE Microwave and Wireless Components Letters*, vol. 11, no. 3, pp. 130-132, Mar. 2001.
- [14] S. K. Davis, E. J. Bond, X. Li, S. C. Hagness, and B. D. Van Veen, "Microwave Imaging via Space-Time Beamforming for Early Detection of Breast Cancer: Beamformer Design in the Frequency Domain," *Journal of Electromagnetic Waves and Applications*, vol. 17, no. 2, pp. 357-381, 2003.
- [15] E. J. Bond, X. Li, S. C. Hagness, and B. D. Van Veen, "Microwave Imaging via Space-Time Beamforming for Early Detection of Breast Cancer," *IEEE Transactions on Antennas and Propagation*, vol. 51, no. 8, pp. 1690-1705, Aug. 2003.

- [16] X. Li, S. K. Davis, S. C. Hagness, D. W. van der Weide, and B. D. Van Veen, "Microwave Imaging via Space-Time Beamforming: Experimental Investigation of Tumor Detection in Multi-Layer Breast Phantoms," *IEEE Transactions on Microwave Theory and Techniques*, vol. 52, no. 8, pp. 1856-1865, Aug. 2004.
- [17] X. Li, E. J. Bond, B. D. Van Veen, and S. C. Hagness, "An overview of ultrawideband microwave imaging via space-time beamforming for early-stage breast cancer detection," *IEEE Antennas and Propagation Magazine*, vol. 47, no. 1, pp. 19-34, Feb. 2005.
- [18] S. K. Davis, H. Tandradinata, S. C. Hagness, and B. D. Van Veen, "Ultrawideband microwave breast cancer detection: A detection-theoretic approach using the generalized likelihood ratio test," *IEEE Transactions on Biomedical Engineering*, vol. 52, no. 7, pp. 1237-1250, July 2005.
- [19] E. C. Fear, M. A. Stuchly, "Microwave system for breast tumor detection," *IEEE Microwave and Guided Wave Letters*, vol. 9, no. 11, pp. 470-472, Nov. 1999.
- [20] E. C. Fear and M. A. Stuchly, "Microwave Detection of Breast Cancer," *IEEE Transactions on Microwave Theory and Techniques*, vol. 48, no. 11, pp. 1854-1863, Nov. 2000.
- [21] E. C. Fear, X. Li, S. C. Hagness, and M. A. Stuchly, "Confocal microwave imaging for breast cancer detection: Localization of tumors in three dimensions," *IEEE Transactions on Biomedical Engineering*, vol. 49, no. 8, pp. 812-822, Aug. 2002.
- [22] E. C. Fear, J. Sill, and M. A. Stuchly, "Experimental Feasibility Study of Confocal Microwave Imaging for Breast Tumor Detection," *IEEE Transactions on Microwave Theory and Techniques*, vol. 51, no. 3, pp. 887-892, Mar. 2003.
- [23] J. M. Sill, E. C. Fear, "Tissue sensing adaptive radar for breast cancer detection - experimental investigation of simple tumor models," *IEEE Transactions on Microwave Theory and Techniques*, vol. 53, no. 11, pp. 3312-3319, Nov. 2005.
- [24] E.C. Fear, S.C. Hagness, P.M. Meaney, M. Okoniewski and M.A. Stuchly, "Near-field imaging for breast tumor detection," *IEEE Microwave Magazine*, vol. 3, pp. 48-56, Mar. 2002.
- [25] B. Guo, Y. Yang, J. Li, P. Stoica, and R. Wu, "Microwave imaging via adaptive beamforming methods for breast cancer detection," *Journal of Electromagnetic Waves and Applications*, vol. 20, no. 1, pp. 53-63, 2006.
- [26] Y. Xie, B. Guo, L. Xu, J. Li, and P. Stoica, "Multistatic adaptive microwave imaging for early breast cancer detection," *IEEE Transactions on Biomedical Engineering*, vol. 53, no. 8, pp. 1647-1657, Aug. 2006.
- [27] Kosmas, P.; Rappaport, C.M., "Time reversal with the FDTD method for microwave breast cancer detection," *IEEE Transactions on Microwave Theory and Techniques*, vol. 53, no. 7, pp. 2317-2323, July 2005.
- [28] Kosmas, P.; Rappaport, C.M., "A matched-filter FDTD-based time reversal approach for microwave breast cancer detection," *IEEE Transactions on Antennas and Propagation*, vol. 54, no. 4, pp. 1257-1264, Apr. 2006.
- [29] D. W. Winters, E. J. Bond, B. D. Van Veen, and S. C. Hagness, "Estimation of the frequency-dependent average dielectric properties of breast tissue using a time-domain inverse scattering technique," *IEEE Transactions on Antennas and Propagation*, vol. 54, no. 11, pp. 3517-3528, Nov. 2006.
- [30] Y. Chen, E. Gunawan, K. S. Low, S. C. Wang, C. B. Soh, and L. L. Thi, "Time of arrival data fusion method for two-dimensional ultrawideband breast cancer detection," *IEEE Transactions on Antennas and Propagation*, vol. 55, no. 10, Oct. 2007.

-
- [31] T. C. Williams, E. C. Fear, D. T. Westwick, "Tissue sensing adaptive radar for breast cancer detection - investigations of an improved skin-sensing method," *IEEE Transactions on Microwave Theory and Techniques*, vol. 54, no. 4, pp. 1308-1314, Jun. 2006.
 - [32] T. C. Williams, J. M. Sill, E. C. Fear, D. T. Westwick, "Breast Surface Estimation for Radar-Based Breast Imaging Systems," *IEEE Transactions on Biomedical Engineering*, vol. 55, no. 6, pp. 1678-1686, Jun. 2008.
 - [33] D. W. Winters, J. D. Shea, E. L. Madsen, G. R. Frank, B. D. Van Veen, S. C. Hagness, "Estimating the Breast Surface Using UWB Microwave Monostatic Backscatter Measurements," *IEEE Transactions on Biomedical Engineering*, vol. 55, no. 1, pp. 247-256, Jan. 2008.
 - [34] W. Zhi, F. Chin, "Entropy-Based Time Window for Artifact Removal in UWB Imaging of Breast Cancer Detection," *IEEE Signal Processing Letters*, vol. 13, no. 10, pp. 585-588, Oct. 2006.
 - [35] F. Buckley and A. Maryott, "Tables of dielectric dispersion data for pure liquids and dilute solutions," National Bureau Standards, *National Bureau Standards Circular*, 589, 1958.
 - [36] S. S. Chaudhary, R. K. Mishra, A. Swarup, and J. M. Thomas, "Dielectric properties of normal and malignant human breast tissues at radiowave and microwave frequencies," *Indian Journal of Biochemistry & Biophysics*, vol. 21, pp. 76-79, 1984.
 - [37] A. J. Surowiec, S. S. Stuchly, J. R. Barr, and A. Swarup, "Dielectric properties of breast carcinoma and the surrounding tissues," *IEEE Transactions on Biomedical Engineering*, vol. 35, no. 4, pp. 257-263, 1988.
 - [38] W. T. Joines, Y. Z. Dhenxing, and R. L. Jirtle, "The measured electrical properties of normal and malignant human tissues from 50 to 900 MHz," *Medical Physics*, vol. 21, pp. 547-550, 1994.
 - [39] M. Lazebnik, E. L. Madsen, G. R. Frank, and S. C. Hagness, "Tissue-mimicking phantom materials for narrowband and ultrawideband microwave applications," *Physics in Medicine and Biology*, vol. 50, pp. 4245-4258, 2005.
 - [40] A. W. Guy, "Analysis of electromagnetic fields induced in biological tissue by thermographic studies on equivalent phantom models," *IEEE Transactions on Microwave Theory and Techniques*, vol. 19, pp. 189-217, 1971.
 - [41] M. G. Bini, A. Ignesti, L. Millanta, R. Olmi, N. Rubino, Vanni, "The Polyacrylamide as a Phantom Material for Electromagnetic Hyperthermia Studies," *IEEE Transactions on Biomedical Engineering*, vol. 31 pp. 317-322, 1984
 - [42] J. J. W. Lagendijk and P. Nilsson, "Hyperthermia dough: A fat and bone equivalent phantom to test microwave/radio frequency hyperthermia heating systems," *Physics in Medicine and Biology*, vol. 30, no. 7, pp. 709-712, 1985.
 - [43] G. Bindu, Anil Lonappan, Vinu Thomas, V. Hamsakkutty, C. K. Aanandan, K. T. Mathew, "Microwave characterization of breast-phantom materials," *Microwave and Optical Technology Letters*, vol. 43, no. 6, pp. 506-508, 2004.
 - [44] G. Bindu, A. Lonappan, V. Thomas, C. K. Aanandan, and K. T. Mathew, "Dielectric studies of corn syrup for applications in microwave breast imaging," *Progress In Electromagnetics Research*, PIER 59, 175-186, 2006.
 - [45] E. L. Madsen, J. A. Zagzebski, G. R. Frank, "Oil-in-gelatin dispersions for use as ultrasonically tissue-mimicking materials," *Ultrasound in Medicine and Biology*, vol. 8, pp. 277-287, 1982.

- [46] E. L. Madsen, J. A. Zagzebski, G. R. Frank, "An anthropomorphic ultrasound breast phantom containing intermediate-sized scatterers," *Ultrasound in Medicine and Biology*, vol. 8, pp. 381-392, 1982.
- [47] E. L. Madsen, J. A. Zagzebski, G. R. Frank, J. F. Greenleaf, P. L. Carson, "Anthropomorphic breast phantoms for assessing ultrasonic imaging system performance and for training ultrasonographers," *Journal of Clinical Ultrasound*, vol. 10, pp. 67-75, 1982.
- [48] M. Lazebnik, L. McCartney, D. Popovic, C. B. Watkins, M. J. Lindstrom, J. Harter, S. Sewall, A. Magliocco, J. H. Booske, M. Okoniewski, and S. C. Hagness, "A large-scale study of the ultrawideband microwave dielectric properties of normal breast tissue obtained from reduction surgeries," *Physics in Medicine and Biology*, vol. 52, pp. 2637–2656, May 2007.
- [49] M. Lazebnik, D. Popovic, L. McCartney, C. B. Watkins, M. J. Lindstrom, J. Harter, S. Sewall, T. Ogilvie, A. Magliocco, T. M. Breslin, W. Tempe, D. Mew, J. H. Booske, M. Okoniewski, and S. C. Hagness, "A large-scale study of the ultrawideband microwave dielectric properties of normal, benign, and malignant breast tissues obtained from cancer surgeries," *Physics in Medicine and Biology*, vol. 52, pp. 6093–6115, Oct. 2007.

Appendices

Appendix A1: Labview Graphic User Interface for Controlling XY-Rotary Experimental Stage and Agilent DSO-081204B Real Time Oscilloscope.

Appendix A2: Labview Program Overall Block Diagram

Appendix A3: Labview Program for Initializing Real-Time Oscilloscope

Appendix A4: Labview Program for Setting Averaging Number

Appendix A5: Labview Program for Setting Measurement Channel

Appendix A6: Labview Program for Setting Time Scale

Appendix A7: Labview Program to Reset Averaging Function

Appendix A8: Labview Program for Setting Voltage Scale

Appendix A9: Labview Program for Setting Triggering Channel and Level

Appendix A10: Labview Program for Setting Interpolation Function

Appendix A11: Labview Program for Controlling Experimental Stage

Appendix A12: Labview Program for Controlling Data Saving

Appendix B1: Galil Motion Controller Program to Reset the Experimental Stage

```
#HOME
SH XYZ
SP 10000,10000,12000
PR -1000,-1000,-1200
BG XYZ
AM XYZ
FI XY
BG XY
AM XY
HM Z
BG Z
AM Z
WT 3000
PR 133000,100000,120000
BG XYZ
AM XYZ
AC 1024,1024,1024
DC 1024,1024,1024
SP 1024,1024,1024
DP,,11
EN
```

Appendix B1: Galil Motion Controller Program to Rotate the Experimental Stage

```
#ROTATE
PR,,600
BGZ
AMZ
WT 10
EN
```

Appendix C1: Matlab Program for Pulse Instability Compensation

```

%-----Pulse Instability Compensation-----%

% original sampling rate is 640e9
sr=5000;      % sampling rate
rr=360;       % antenna number
XX=zeros(ceil(8177*sr/640),rr);
XX=resample(data',sr,640);

% normalize the peak-peak amplitude to one
NX=sum(XX,2)/rr;
h=max(NX)-min(NX);
NX=NX/h;

% define alignment position 640*1.5625=1000
where=ceil(sr*1.5625);

% finding the zero crossing point btw peaks
[p p1]=min(NX);
[p p2]=max(NX);
p=abs(p1-p2);
p1=min([p1 p2]);
p2=p1+p;
[p p3]=min(abs(NX(p1:p2)));
p3=p3+p1;
p=where-p3;

% shifting all signals to defined position
for i=1:rr
    [p p3]=min(abs(XX(p1:p2,i)));
    p3=p3+p1;
    p=where-p3;
    h=max(XX(:,i))-min(XX(:,i));
    XX(:,i)=circshift(XX(:,i),p)/h;
    hh=min(XX(:,i))-min(NX);
    XX(:,i)=XX(:,i)-hh;
End

% resample to original sampling rate
X=resample(XX,640,sr);

%%-----end-----%%

```

Appendix C2: Matlab Program for Delay & Sum Beamforming Part 1

```

%-----Delay & Sum Beamforming 1-----%

% make room for large variables
ener=zeros(201,201);
sss=zeros(401,201,201);
dd=zeros(201,201,360);
dt=zeros(201,201,360);

q=50;                % breast phantom size
sr=640e+9;           % sampling rate
sp=0.9e+8;           % speed of light in breast
for i=1:360           % antenna number
    deg=i;

    % xy position of transmit & receive antenna pair
    ix1=101+q*cosd(deg+15);
    iy1=101-q*sind(deg+15);
    ix2=101+q*cosd(deg-15);
    iy2=101-q*sind(deg-15);

    % calculating with image of 201x201 pixels
    for y=1:201
        for x=1:201

            % calculating round region within the breast radius
            r=sqrt((y-101)^2+(x-101)^2);
            if r<=q
                dx=x-ix1;
                dy=y-iy1;
                d1=sqrt(dx^2+dy^2);
                dx=x-ix2;
                dy=y-iy2;
                d2=sqrt(dx^2+dy^2);
                dd(y,x,i)=d1+d2;

            % calculating round trip delay time from antennas
            dt(y,x,i)=round(+dd(y,x,i)*0.001/sp*sr);
            end
        end
    end
end
end

```

Appendix C3: Matlab Program for Delay & Sum Beamforming Part 2

```

%-----Delay & Sum Beamforming 2-----%

% setting the signal window length
p2=1000;
p1=p2;
p2=p2+400;

% summing the windowed signals
for y=1:201
for x=1:201
    r=sqrt((y-101)^2+(x-101)^2);
    if r<=q
        for i=1:360
            wind=p1+dt(y,x,i):p2+dt(y,x,i);
            ss(:,i)=s(wind,i);
        end
        sss(:,x,y)=sum(ss(:,,:),2);
        ener(y,x)=var(sss(:,x,y));
    end
end
end

% plotting signal energy for breast image
figure();
[m x0]=max(max(ener));
[m y0]=max(ener(:,x0));
imagesc(ener); colorbar;

m
xlim([20 180]); ylim([20 180]);

%%-----end-----%%

```


Appendix C4: Matlab Program for Checking Phase Shifted Signal

```

%-----Checking Phase Shifted Signal-----%

% Define location to check
x=x0; y=y0;
% x=131; y=101;

% Shifting the signals for defined location
figure();
for i=1:360
    deg=i;
    ix1=101+q*cosd(deg+15);
    iy1=101-q*sind(deg+15);
    ix2=101+q*cosd(deg-15);
    iy2=101-q*sind(deg-15);
    dx=x-ix1;
    dy=y-iy1;
    d1=sqrt(dx^2+dy^2);
    dx=x-ix2;
    dy=y-iy2;
    d2=sqrt(dx^2+dy^2);
    dd(i)=d1+d2;
    dt(i)=round(-(dd(i)-0)*0.001/sp*sr);
    ss(:,i)=circshift(s(:,i),dt(i));
end

% Plotting shifted signal
subplot(3,1,1); plot(ss(1:4000,:));

% Plotting windowed signal
sss=sum(ss(:,:),2);
subplot(3,1,2); plot(sss(1:4000)); hold;
t=1:8001;
plot(t(p1:p2),sss(p1:p2),'LineWidth',2);
plot(N(1:4000),'r'); hold;

% Plotting power spectrum
ff=fft(sss);
psd=ff.*conj(ff)/8001;
f=(0:8000)*640e9/8001;
subplot(3,1,3); plot(f(1:240),psd(1:240));

%%-----end-----%%

```

Appendix C5: Matlab Program for Calculating Image SNR

```

%-----Image SNR-----%

ene1=zeros(201,201);
ene2=zeros(201,201);

for y=1:201
for x=1:201
    r=sqrt((y-101)^2+(x-101)^2);
    if r<=q
        r0=sqrt((y-131)^2+(x-101)^2);

% Define the tumor proximity
% Medium: air 12, oil 8

        if r0>=12
            ene1(y,x)=ener(y,x);
        else
            ene2(y,x)=ener(y,x);
        end
    end
end
end

% maximum clutter pixel energy

n1=max(max(ene1));
n2=max(max(ene2));
SNratio=10*log10(n2/n1)

clear x y;

%%-----end-----%%

```

Appendix C6: Matlab Program for Interference Filtering

```

%-----Filtering-----%

load bandstop;
f=(0:6400)*640e9/6400;
for i=1:rr
    ff=fft(s(:,i),6400);
    psd(:,i)=ff.*conj(ff)/6400;
end

% Performing filtering
c0=0; c1=0; c2=0;
for i=1:rr
    psdd=psd(19,i);
    if psdd>1e-3
        s1=conv(s(:,i),Num18);
        cv=ceil(length(Num18)/2);
        s2(:,i)=s1(cv:8000+cv);
        c1=c1+1;
    else
        s2(:,i)=s(:,i);
    end
    psdd=max(psd(25:26,i));
    if psdd>1e-3
        s3=conv(s2(:,i),Num24);
        cv=ceil(length(Num24)/2);
        s4(:,i)=s3(cv:8000+cv);
        c2=c2+1;
    else
        s4(:,i)=s2(:,i);
    end
    psdd=max(psd(1:24,i));
    if psdd>1e-3
        s5=conv(s4(:,i),Num0);
        cv=ceil(length(Num0)/2);
        s6(:,i)=s5(cv:8000+cv);
        c0=c0+1;
    else
        s6(:,i)=s4(:,i);
    end
end

s1=s6; s0=s;
for i=1:rr
    ff=fft(s1(:,i),6400);
    psd(:,i)=ff.*conj(ff)/6400;
end

%%-----end-----%%

```

Appendix C7: Matlab Program for Correlation

```

%-----Correlation-----%

load template;

% delay & sum beamforming
for y=1:201
for x=1:201
    r=sqrt((y-101)^2+(x-101)^2);
    if r<=q
        for i=1:360
            wind=p1+dt(y,x,i):p2+dt(y,x,i);
            ss(:,i)=s(wind,i);

            %ss(:,i)=ss(:,i)*(dd(y,x,i)/50)^2;
            %apply radial spreading compensation
        end
        sss(:,x,y)=sum(ss(:,,:),2);
        ener1(y,x)=var(sss(:,x,y));
        ener2(y,x)=sum(temp.*sss(:,x,y));
        %correlate with ideal tumor response
    end
end
end

% shifting all pixels to positive
ener2=ener2-min(min(ener2));

% multiplying with delay&sum image
ener =ener1.*ener2;

% plotting image
figure();

[m x0]=max(max(ener));
[m y0]=max(ener(:,x0)); m
imagesc(ener); colorbar;

%%-----end-----%%

```

Appendix C8: Matlab Program to Generate Gaussian Pulse

```

%-----Gaussian Pulse Spectrum-----%

f0=6e9;           %6GHz
t0=8e-11;         %80ps
sr=4e+10;         %40GSa/s
t=0:1/sr:1e-6;   %1us

% Sine Wave modulated Gaussian Pulse
G=1*(exp(-(t-4*t0).^2)/t0^2));
S=1*sin(2*pi*f0.*(t-4*t0));
P=1.*G.*S;

subplot(2,3,1); plot(t(1:50),S(1:50));
xlabel('Time (nsec)');
ylabel('Amp (Volts)');
subplot(2,3,2); plot(t(1:50),G(1:50));
subplot(2,3,3); plot(t(1:50),P(1:50));

% Calculating Power Spectrum from FFT
ws=40000; %FFT window of 1us
freq=1:ws;
freq=sr/ws*(freq-1);
ff=fft(S,ws);
PSD=ff.*conj(ff)/ws;
dBm=10*log10(PSD./ws*2/50/0.001);
%divide by 50ohm 1mW, time 2-side FFT

subplot(2,3,4);
plot(freq(1:ws/4),dBm(1:ws/4));
xlabel('Frequency (GHz)');
ylabel('PSD (dBm / MHz)');

ener=sum(G(1:ws).^2);
rms=sqrt(ener/ws);
acpwr=rms^2/50;
dBm=10*log10(sum(PSD)/ws/50/0.001);

ff=fft(P,ws);
PSD=ff.*conj(ff)/ws;
dBm=10*log10(PSD./ws*2/50/0.001);
subplot(2,3,6);
plot(freq(1:ws/4),dBm(1:ws/4));
ylim([-100 -40]);

%%-----end-----%%

```

Appendix C9: Matlab Program to Perform Inverse FFT on S21 data

```

%-----Inverse Fourier Transform-----%

% Loading Gaussian Pulse Spectrum
load PSD2;

% Loading S-parameter file from VNA
for l=1:24
    Name=strcat(int2str(l),'.s2p');
    s=dlmread(Name,'',[6 3 206 4]);
    si1=s(:,2);
    ss1=10.^(s(:,1)/20).*PSD;

% Creating complex conjugate part
    for i=1:201
        ss2(202-i,:)=ss1(i);
        si2(202-i,:)=si1(i);
    end
    ss=[ss1;ss2];
    si=[si1;-si2];
    sx=ss.*cosd(-si);
    sy=ss.*sind(-si);
    sz=sy+j*sx;
    ff=ifft(sz,'symmetric');
    P(:,l)=ff;
end

t=(1:9529)/640e9;

Y=resample(P,640,27);

figure(); plot(t,Y);

%%-----end-----%%

```

Appendix C10: Matlab Program to Calculate Noise Level

```

%-----Noise Level-----%

rr=360;                %antenna number
le=8001;               %signal length
s=S(1:le,:);
t=(1:le)/640e9;        %time vector
f=(0:le-1)*640e9/le;   %freq vector

figure();

% computing sum of square
for i=1:rr
    e(:,i)=sum(s(:,i).^2);
    ff=fft(s(:,i));
    psd(:,i)=ff.*conj(ff)/le;
end
    subplot(3,1,1);
    plot(t(1:le),s(1:le,:));
    xlim([0 10e-9]); ylim([-0.005 0.005]);

    subplot(3,1,2);
    semilogy(f(1:240),psd(1:240,:));
    ylim([1e-10 1]);

% computing averaged RMS amplitude
e=sqrt(e./le);
mean(e)
subplot(3,1,3); plot(e);
ylim([0 0.002]);

% computing averaged Peak-Peak amplitude
for i=1:rr
    for j=1:(le-1)/1000
        m(j,i)=max(s((j-1)*1000+1:j*1000,i))-min(s((j-1)*1000+1:j*1000,i));
    end
end
mean(mean(m))

%%-----end-----%%

```