

Improving Shell-Based Phantom Materials and Quality Assurance in Breast Microwave Sensing

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Abstract

Breast Microwave Sensing (BMS) systems require adequate testing and performance before reaching clinical trials. This thesis presents improved materials for shell-based breast phantoms and quality assurance devices to provide comparable testing in the field. Previous studies utilized 3D-printable plastics to create shells for shell-based phantoms; however, the low permittivity of these plastics impacts the phantoms' dielectric accuracy. The liquids filling shell-based phantoms also often contain air bubbles, leading to undesirable microwave scattering. This research explores new tissue-mimicking materials to overcome these issues. The low-permittivity 3D-printed plastic filament have been substituted with graphite, carbon black, and resin to replicate the skin's properties within the 0.4-9.0 GHz range. Diethylene Glycol Butyl Ether (DGBE) solutions have replaced glycerin and Triton X-100 to simulate the characteristics of adipose and fibroglandular tissue. The resin-based material more accurately mimicked the properties of *ex vivo* tissue samples than 3D-printed plastics. DGBE solutions demonstrated superior dielectric properties compared to glycerin and Triton X-100 solutions. These DGBE solutions are preferable over glycerin and Triton X-100 due to their lower viscosity, reduced tendency for air bubble formation, better short-term stability, temperature stability, and enhanced long-term stability, which supports the reusability of these materials. Quality assurance devices were designed and 3D-printed to analyze and compare BMS systems, an area the field has been lacking in the literature. The materials studied in this work improve on existing breast phantoms by ensuring dielectric accuracy and stability, offering better experimental utility for microwave breast imaging. At the same time, the devices presented can be used for quality metrics to appropriately compare BMS systems on a fundamental level.

Contributions

The following are the primary contributions presented by the author in this thesis:

- A review of Breast Microwave Sensing (BMS) phantoms, the definition of an ideal breast phantom, and a proposition of how the ideal breast phantom could be achieved.
- Improvements in the design and fabrication of a free-space characterization device.
- Determination of the anisotropy of carbon-doped 3D-printable filaments in S -parameter measurements.
- Determined the feasibility of using carbon resin to replace low-permittivity plastics in fabricating shell-based phantom shells to allow for more realistic responses.
- Measured the dielectric properties of Diethylene Glycol Butyl Ether (DGBE) to be used in BMS experimentation. This includes mathematical modelling of Cole-Cole parameters as a function of DGBE concentration, determining adipose and fibroglandular mimicking solutions, and determining the thermal and temporal stability of the solutions.
- Creation of a phantom positioning device, allowing more controllable experimentation in BMS
- Collection of a point-like source data set at known positions. A regression methodology used the data set to correct positioning errors and minimize the uncertainty in positioning for a BMS system.
- Design of a spatial resolution device and cylindrical phantom for noise characterization, including data collection for the metric determination.

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- Determination of suitable containers for contrast resolution measurements using different DGBE concentrations.

Acronyms

AAPM American Association of Physicists in Medicine. i, 68

ABS Acrylonitrile Butadiene Styrene. i, 19, 20

AUC Area Under the Curve. i, x, xi, xii, 34, 39, 40, 41, 42, 92

BD Breast Density. i, 2

BI-RADS Breast Information Reporting and Data Systems. i, 2

BMS Breast Microwave Sensing. i, ii, vii, x, xi, xiii, 1, 3, 4, 5, 6, 7, 8, 9, 10, 11, 12, 19, 20, 21, 23, 31, 42, 48, 49, 65, 67, 68, 69, 70, 71, 72, 76, 77, 83, 85, 88, 89, 90, 91, 92, 94, 95

CB Carbon Black. i, xii, 21, 35, 43, 46, 47, 93

CF-PLA Carbon Fibre Polylactic acid. i, xi, xii, 31, 32, 34, 39, 40, 42, 44, 45, 92

CoM Center of Mass. i, 74

CPLA Conductive Polylactic acid. i, xii, 31, 32, 34, 39, 41, 42, 44, 45, 46, 48, 92

CT Computed Tomography. i, 21, 35

DAS Delay-and-Sum. i, 7, 74, 81, 88, 94

DGBE Diethylene Glycol Butyl Ether. i, ii, iii, viii, x, xii, xiii, 50, 51, 52, 53, 54, 55, 56, 57, 58, 60, 61, 62, 63, 64, 65, 66, 77, 78, 85, 89, 93, 94, 95

DMAS Delay-and-Sum. i, 7

FD Frequency-Domain. i, 74

FDM Fused Deposition Modelling. i, viii, 31, 35, 39, 42, 48, 49, 87, 89, 92

FF Far-Field. i, 28

FP False-Positive. i, 2, 3, 8, 9

ICZT Inverse Chirp-Z Transform. i, 74

LE Localization Error. i, 67, 68, 81, 82, 89, 94

MAPE Mean Absolute Percent Error. i, xii, xiii, 43, 46, 48, 53, 56, 57, 62, 63, 65, 66, 93, 94

ML Machine Learning. i, 5, 7, 8, 9, 91

MRI Magnetic Resonance Imaging. i, 2, 3, 9, 21, 22, 35

MUT Material Under Test. i, x, xi, xii, 28, 30, 33, 34, 37, 38, 40, 41, 48

OSL open-short-load. i, 38

OUT Object Under Test. i, 68, 73, 75, 77, 89, 96, 97

PC-CPE Polycarbonate Co-polyester. i, xiv, 31, 34, 35, 46, 77, 78, 85, 87, 89, 93, 95

PLA Polylactic acid. i, 25, 31, 73, 83

PSF Point Spread Function. i

QA Quality Assurance. i, 68, 88, 89

SLA Stereolithography. i, xiv, 77, 85, 87, 89, 95

SNR Signal to Noise Ratio. i, x, 78, 88, 89, 95

TD Time-Domain. i, xiii, xiv, 74, 81, 82, 83, 84, 88

TMMs Tissue Mimicking Materials. i, x, 5, 11, 18, 21, 25, 35, 38, 43, 46, 47, 50, 51, 64, 65

VNA Vector Network Analyser. i, 28, 33, 34, 38, 69, 71, 82

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1. Introduction to Breast Microwave Sensing

Breast Microwave Sensing (BMS) will be presented as an alternative breast cancer screening modality in this chapter. Current breast cancer detection methods, introduction to BMS, and the need for large, diverse, and accurate datasets will be discussed.

1.1 Background and Justification

In 2020, breast cancer was projected to be the second most diagnosed cancer in Canada and the most common among women; furthermore, it is projected to have the fourth highest number of deaths of all cancers, both male and female [1]. The Canadian Cancer Society estimates that 1 in 8 Canadian women will be diagnosed with breast cancer in their lifetime, and 1 in 33 will die from breast cancer [2]. Worldwide, breast cancer has surpassed lung cancer as the most prevalent cancer, with less-developed regions having a higher mortality rate [3]. The ideal system should be sensitive, accurate, robust, inexpensive, and user-friendly to detect early-stage breast cancer in the developing world. Such a system will facilitate treatment when the tumour is still small, thereby reducing mortality while making it possible to deploy the device to countries where the limited human and capital infrastructure makes existing X-ray-based breast screening difficult. However, more evidence is required for the benefits of screening under the age of 50 and over the age of 70 [4].

The current standard in breast cancer screening is X-ray mammography. During a mammogram, ionizing X-ray radiation irradiates the breast. Different tissues will attenuate different

amounts of radiation, allowing various tissues within the breast to be imaged. However, the benefits of using mammography for screening have been questioned [5]. The use of ionizing radiation will deliver a dose to the healthy tissue in the breast, which can damage the healthy tissue. In Canada and the United States of America, it is recommended that women get mammograms from 40-74 years of age as there is evidence for a moderate reduction in deaths caused by breast cancer [4,6].

Within the breast, healthy tissues can be grouped into two types: adipose tissue, or fatty tissue, and fibroglandular tissue, or dense tissue. The Breast Density (BD) is given by the relative amount of fibroglandular tissue to adipose tissue; the American College of Radiology has created four classifications of BD in the Breast Information Reporting and Data Systems (BI-RADS); 1) almost entirely fatty, 2) fatty with scattered regions of fibroglandular, 3) heterogeneously dense, 4) extremely dense [7]. Breast cancer is not only more likely, but the sensitivity of mammography also drops from about 88.2% in fatty breasts to 62.2% in dense breasts [8–10]; this means that dense tissue results in a tumour that is harder to detect. Furthermore, approximately 43% of women between 40-74 are in BI-RADS classes 3 and 4 [11]. Even more women in the 40-49 age range have a higher BD [11,12], meaning that in younger women, it is even harder to detect breast cancer and more likely to have a False-Positive (FP) result.

The relatively high FP rate is also a problem in mammography. The probability of a FP for a single mammogram is approximately 10%, with the FP rate improving with the age group of women [13–15]. Furthermore, in ten years, the cumulative probability of a FP is 56.3% (49.6%) for digital mammography (tomosynthesis) or 38.1% (35.7%) for biennial screening [14]. The impacts of the high FP rate are negative; women whose mammograms resulted in an FP are reported to have more anxiety and stress related to the problem [16]. The result of the FP does not only affect the mentality of the patient but can also have financial and potentially physical costs; patients will be required to undergo further screening, such as additional mammography, ultrasound, Magnetic Resonance Imaging (MRI), or even physically invasive biopsy [14].

Other imaging modalities exist that can detect breast cancer. Ultrasound uses high-frequency sound waves to image internal tissues utilizing the contrast in the acoustic impedances of different tissues; this is specifically helpful in women with more dense breasts, where mammography is less effective [17]. Despite the use of non-ionizing radiation and the ability to detect cancer in denser breasts, sonography requires the use of highly skilled operators, which may cause variability in the results [17]. Furthermore, sonography is often used as an adjunct to mammography, which may increase the sensitivity and the FP rate, leading to questions of whether benefits outweigh harms [17–19].

MRI is another non-ionizing imaging modality capable of detecting breast cancer. MRI, like ultrasound, has the potential to improve screening in high-density breasts [9], has good sensitivity but lacks specificity, increasing the likelihood of FP results, and has increased costs in comparison to ultrasound and mammography [20,21], making it unsuitable for screening the general population.

1.2 An Introduction to BMS

Breast Microwave Sensing (BMS) has been an active area of research over the past forty years as a potential alternative to current methods of breast cancer screening, strongly motivated by the use of non-ionizing radiation, lack of breast compression, and relatively low cost in comparison to other modalities [22–25]. These BMS systems typically operate between frequencies of 0.3–10 GHz [22–24], and their capabilities are dependent on the scatter caused by the contrast in the complex permittivity of the tissue, $\hat{\epsilon}$, given by

$$\hat{\epsilon} = \epsilon' - j\epsilon'', \quad (1.1)$$

where ϵ' and ϵ'' are the real and imaginary parts of the complex permittivity, respectively.

However, it is more commonly stated in terms of the relative permittivity, $\hat{\epsilon}_r$, given by

$$\hat{\epsilon}_r = \frac{\hat{\epsilon}}{\epsilon_0} = \frac{1}{\epsilon_0}(\epsilon'_r - j\epsilon''_r), \quad (1.2)$$

where ϵ_0 is the permittivity of free space. For the remainder of this work, the real part will be denoted ϵ_r , and the imaginary part will be in terms of conductivity, σ , allowing us to denote the complex relative permittivity as

$$\hat{\epsilon}_r = \frac{1}{\epsilon_0} \left(\epsilon_r - j \frac{\sigma}{\omega} \right), \quad (1.3)$$

where ω is the angular frequency, or $2\pi f$, where f is the frequency. The primary source of scattering is due to the contrast in the dielectric properties between two materials; in BMS, ϵ_r is a key factor that influences the scattering characteristics of the breast.

1.2.1 Dielectric Properties of the Breast

For BMS, the tissues of importance are skin, adipose or fatty, fibroglandular or dense, and unhealthy or cancerous. Even more critical are the dielectric properties of the tissues at the frequencies used, as this contrast will cause the scatter that will allow for tumour detection [26]. The contrast is based on the water content of that tissue, and specifically for unhealthy tissue, the amount of blood drawn. Blood is approximately 60% plasma, and plasma has a high water content (approximately 92% by volume) [26, 27]. Early studies of the dielectric properties of breast tissue suggested that the contrast between healthy and unhealthy tissues was high, as high as 10:1, leading to the promising potential of BMS [28–30]. However, more recent studies have shown that contrast can be significantly lower for tumours in high-density tissues (approximately 1.25:1–2:1) [31–35].

The dielectric properties of adipose, fibroglandular, and tumour tissue can be modelled by a modified two-pole Cole-Cole model [34] given by

$$\epsilon_r(\omega) = \epsilon_\infty + \frac{\epsilon_m - \epsilon_\infty}{1 + (j\omega\tau_Q)^{1-\beta}} + \frac{\epsilon_s - \epsilon_m}{1 + (j\omega\tau_p)} - \frac{\sigma_s}{\omega\epsilon_0}j, \quad (1.4)$$

where ω is the angular frequency, σ_s is the effect conductivity, ϵ_s , ϵ_m , ϵ_∞ are permittivities of low-, middle-, and high-frequency limits, τ_p and τ_Q are the relaxation times at low and high frequencies, and β is a measure of broadening of the dispersion. Skin tissue can similarly be modelled by a regular two-pole Cole-Cole model [30], given by

$$\epsilon_r(\omega) = \epsilon_\infty + \frac{\epsilon_m - \epsilon_\infty}{1 + (j\omega\tau_Q)^{1-\beta}} + \frac{\epsilon_s - \epsilon_m}{1 + (j\omega\tau_p)^{1-\beta}} + \frac{\sigma_s}{j\omega\epsilon_0}. \quad (1.5)$$

The models can be used to determine the properties of the breast at frequencies of interest. These modelled properties can be used for multiple purposes to study the uses of microwave technology for breast cancer detection, the most important being used in simulations or for comparison when trying to determine Tissue Mimicking Materials (TMMs). A plot of tissues modelled using the respective models from Equation 1.4 and Equation 1.5 over the frequency range typically used for BMS, i.e., 0.3–10 GHz, is shown in Figure 1.1

1.3 Current Detection and Testing Methods

The current detection methods in BMS include image-based and Machine Learning (ML). Image reconstruction aims to create a spatial profile of tissue's dielectric properties or reflections, allowing cancerous tissue detection. There are two imaging techniques: tomography and radar imaging [36]. On the other hand, ML techniques are being researched as an alternative to image-

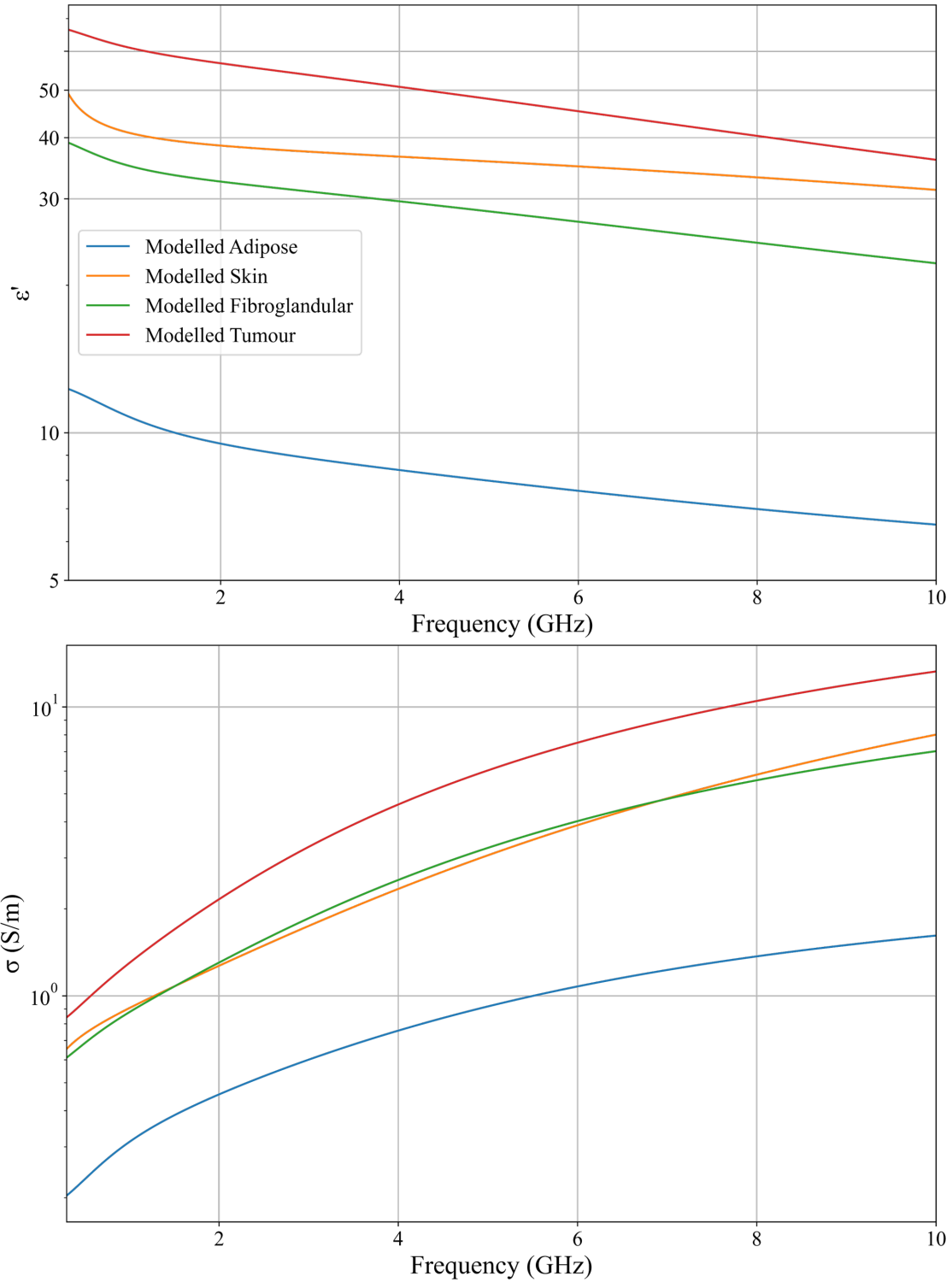


Figure 1.1: The modelled permittivity and conductivity of tissues of interest in BMS over the frequency range of 0.3–10 GHz.

based methods [37,38]. ML allows for the objective analysis of complex data and provides automation and quick diagnostic tasks, removing the potential for human error. Both detection methods need controlled testing to determine if they are sufficient for breast cancer detection; data is needed to learn, improve, and test current methods.

1.3.1 Image-Based Detection

Image-based detection in BMS uses signal processing to reconstruct breast images from raw data. Tomography-based methods model the physics of microwave propagation well by solving a non-linear, inverse scattering problem iteratively to recover a profile of the dielectric properties of the breast [39]. Since the algorithms iteratively simulate the propagation of microwaves through the modelled breast until the simulated signal matches the measured signal, this method is computationally expensive.

Alternatively, radar-based methods utilize scattered signals to create a breast reflectivity map to reveal the presence or absence of a tumour [40, 41]. The Delay-and-Sum (DAS) was the first radar-based algorithm proposed and, to this day, is the standard algorithm used [26]. The DAS algorithm will time align the signals from all antennas to a spatial position, $\mathbf{r}$, then sum the signals, given mathematically by

$$I(\mathbf{r}) = \sum_{m=1}^M s_m(t_m(\mathbf{r})), \quad (1.6)$$

where s_m is the signal from the m^{th} antenna position, and $t_m(\mathbf{r})$ is the approximate time-of-response corresponding to the spatial position $\mathbf{r}$.

Other beamformers exist, such as the Delay-and-Sum (DMAS) [42], which is a modification of the DAS beamformer with an additional pairwise multiplication procedure after the time alignment to have clutter rejection based on increased sample size. Other algorithm modifications exist [43], including algorithms that model more of the physics behind signal generation [44].

1.3.2 Machine Learning-Based Detection

ML methods use algorithms trained with data to perform a task quickly. The algorithm is given an input and, based on the features of that input, generates an output. These algorithms are typically trained to learn the input-output relationships by using inputs with known outcomes. After the training, the algorithms can be tested with more data with known results as the output to determine how well it performs. For the case of BMS, the input would be some form of the collected data of a scan, while the output could be whether the tumour is benign or malignant [45, 46], or if cancer is present within the breast [37, 47, 48]

1.3.3 Testing of Detection Methods

Numeric breast models, i.e., simulation studies, have been used primarily to test tomographic imaging capabilities [49–52], while the research group at Dartmouth College has investigated the feasibility experimentally [53, 54]. Conversely, radar-based imaging has rapidly transitioned from simulation to experimental testing [39, 55]. Many groups have tested their radar-based algorithms on phantoms [42, 44, 56–63], while others have tested on patients [41, 43, 64–66]. From these studies, it is essential to note that the algorithms can locate tumours in some breasts. However, in others, they can also result in FPs – the algorithms must be thoroughly tested using various breast shapes, sizes, and densities that represent those in the population under investigation.

To train and test ML capabilities, some groups have demonstrated algorithms on simulated datasets [46, 48, 67, 68]. At the same time, most have used experimental phantom measurements [37, 45, 47, 69–73], and another group even showed promising results on patient data [74]. However, in [74], individual antenna measurements were used as input, and training and testing data included scans from the same patient, allowing data contamination, which has been shown to falsely improve the validity of results [37]. The data used to train and test ML algorithms should be large and diverse, but few groups have used datasets of more than 1000 samples [38].

Large and diverse data sets are required to adequately test systems for image- and ML-based detection methods. Before testing on patients, researchers should be able to illustrate a system's capability on experimental breast phantoms. Thus, these phantoms should not only represent the properties and morphology of the breast seen clinically but also represent the variety of breasts seen clinically and be able to create large datasets for ML purposes.

1.4 Problem Definition

1.4.1 Problem Statement

Current imaging modalities for breast cancer detection include x-ray mammography, ultrasound, and MRI. These modalities have been shown to have limitations, such as high FP rates, high cost, large size, and the need for trained personnel. BMS has been studied as a potential alternative to current methods due to the use of non-ionizing microwaves, the relatively low cost, and the potential to reduce the FP rate. However, to adequately test BMS systems and their detection methods, dielectrically, morphologically, and diverse breast phantoms are required. Furthermore, large data sets are needed to train and test these algorithms with the increasing investigation of ML-aided detection in medical imaging.

1.4.2 Research Goals

Hypothesis: Investigation into the materials used in shell-based phantoms can lead to an ideal phantom design, and 3D-printing of breast phantoms can be utilized to improve quality assurance and control in BMS

The goals of this research are:

- Review current breast phantoms and define the ideal breast phantom (Chapter 2).

-
- Investigate solid carbon-doped materials that can replace current low-permittivity plastics used to construct the shell of shell-based phantoms (Chapter 3).
 - Investigate low viscosity liquids with a wide range of dielectric properties that can be used to mimic various tissues in BMS (Chapter 4).
 - Utilization of 3D-printing technologies to improve quality assurance and control in BMS (Chapter 5).

2. A review of Current Breast Microwave Phantoms

The purpose of this chapter is to review the current breast phantoms used in BMS and define the ideal breast phantom.

2.1 Introduction

Clinical trials are the best evaluation for a diagnostic test in development, but they introduce several complex factors. With patient data, the exact tissue geometry and dielectric properties are unknown. Although other imaging modalities can be used to determine tissue geometries, they can introduce further errors, and the dielectric properties are still unknown. Patient data also presents challenges due to its cost, the need for rigorous ethical considerations, and difficulties in interpretation. However, phantom-based work allows for controlled experimentation and testing.

Breast phantoms are models of the breast that facilitate the evaluation of systems and methods in BMS. These tools allow for controlled experiments in the laboratory – the geometry and dielectric properties of the tissue-mimicking components can be well known, providing the opportunity for rigorous analysis of image quality and diagnostic performance. Although phantoms are not perfect breast models, robust and controlled experimentation that can be performed using these tools is necessary and advantageous in developing a novel diagnostic tool.

Three primary categories of breast phantom have been used in BMS: solid-based, gel-based, and shell-based. Solid-based phantoms use solids as TMMs. Gel-based phantoms use semi-solid gels as TMMs. The water content in these phantoms can be adjusted to accurately represent differ-

ent breast tissue types. However, existing fabrication procedures do not allow for precise knowledge of tissue geometries due to the mixing of gels within the phantom. Because of this mixing, each gel-based phantom is typically unique geometrically, complicating the analysis of data collected with gel-based phantoms and limiting the reproducibility of individual phantoms. These phantoms also usually have a shelf life of days or weeks [75].

Solid-based phantoms are generally mechanically robust and stable over time but allow limited configurations of the individual tissue components [76–78]. Furthermore, when the current BMS literature is examined, the fabrication of solid materials mimicking the tumour and fibroglandular has been challenging due to the relatively low permittivity of most solid materials and air bubbles in most moulding agents [76, 79].

Shell-based breast phantoms combine solid and liquid materials to mimic the morphology and dielectric properties of the breast. Solid adipose and fibroglandular shells are used as containers of liquids chosen to mimic the properties of the corresponding breast tissue [60]. These phantoms are uniquely positioned to accurately model tissue morphology and dielectric properties while allowing for complete knowledge of the phantom geometries through 3D-printing for fabrication. The primary challenge when using shell-based phantoms is the dielectric matching of the shell layers: 3D-printing plastics are low-permittivity, do not mimic tissue properties, and may produce unwanted signals [37, 80].

Despite the push for more extensive clinical trials, phantom-based research is better suited to solve the challenges facing BMS as a potential breast cancer detection technique. Phantoms allow for greater control of experimental conditions, *a priori* knowledge of tissue geometries and dielectric properties, and rapid and robust data collection. This review examines the existing methodologies and fabrication techniques in BMS and recommends the design of an ideal phantom.

2.2 The Ideal Breast Phantom

Based on the requirements for testing systems and algorithms, the ideal breast phantom should have the following properties:

- **Dielectrically accurate:** the properties of each tissue represent the average and range of dielectric properties seen clinically.
- **Morphologically correct:** the tissues should represent the morphology of the respective tissue, preferably MRI/CT derived.
- **Stable over time:** the properties should remain stable over a reasonable amount of time, i.e., the phantom should not need to be frequently re-fabricated.
- **Reproducible:** the phantom should be able to be recreated by the same research group, or even better, by other research groups for comparison (open access files if CAD design)
- **Complete knowledge of the tissue locations:** knowledge of where the tissue components, i.e., tumour and fibroglandular tissues, are located within the breast; minimal uncertainty in tissue component positions
- **Versatile and diverse:** the more scans from "one" phantom design, the better mix and match adipose + fibroglandular, freedom of tumour positioning, and freedom of fibroglandular positioning. The phantoms should also be able to represent a range of shapes, sizes, and densities.
- **Low cost:** Ideally, it uses lost-cost materials, especially if dielectric properties are unstable over time.

2.3 Gel-Based Phantoms

Gel-based phantoms have been proposed and modified the most in breast microwave sensing [56, 75, 81–102], the first being developed in 2008. One group compared using Triton X-100 mixtures and oil-in-gelatin mixtures [75], a different group used an outer shell to contain the gelatin-based inner tissues [56], and another used a breast shell used in ultrasound training to contain the gel-based material [102]. One paper presented a phantom with the dielectric properties of the breast that may also be used for studies of breast hyperthermia [93]. More recent advances in gel-based phantoms include the development of multimodal phantoms [91, 99], and phantoms that can be imaged using other modalities such as CT, ultrasound, or MRI.

2.3.1 Composition and Fabrication

The benefit of using gel-based materials is that they allow for the use of low-cost, non-toxic, easy-to-obtain materials capable of mimicking the dielectric properties of the breast. The composition of gel-based phantoms may be traced back to using kerosene and safflower oil-in-gelatin materials to mimic breast tissues in 2005 [103]. While the compositions of gel-based phantoms have varied across the various groups, the majority of groups use the oil-in-gelatin mixtures consisting of a base with oil, gelatin, and water [75, 81, 84, 85, 87, 88, 92–97, 102]. However, the oil used typically varies; other chemicals are occasionally used in these solutions. For the groups that did not use a gelatin base, an agar base was used [83, 90, 91, 99, 100], with one group using a TX151, polythene powder, and water mixture [56], and two groups using both gelatin and agar [82, 95]. Another uses an agar-gelatin powder base [86, 89], and one of the groups uses vinegar as a preservative [86]. One group also used alcohol as the skin-mimicking material [81].

Oil-in-Gelatin Ingredients

Some of the common ingredients used in oil-in-gelatin phantoms are:

-
- Gelatin
 - Oil (safflower, sunflower, grapeseed, etc.)
 - Deionized water
 - Salt
 - Propylene glycol
 - Formaldehyde
 - p-toluic acid
 - I-propanol or n-propanol
 - Detergent (surfactant)

Not all ingredients are used together, with most tissues consisting of oil, water, and gelatin, the oil-in-gelatin base. Typically, the phantoms would consist of either the simple oil-in-gelatin, with different percentages of oil dictating the permittivity of the tissue or include one of the ingredients listed above. The process for creating the tissue-specific mixtures may be found in the respective publications.

Agar-based Ingredients

Compared to oil-in-gelatin phantoms, agar-based phantoms have more diversity in the ingredients used. Table 2.1 shows some of the ingredients used in tissues for the various agar or agar-gelatin phantoms. It should be noted that PEP is polyethylene powder and that both multimodal phantoms are agar-based.

Table 2.1: Table of some of the agar-based phantom ingredients used

[82]	[83]	[89]	[90]	[91] (multi-modal)	[95]
Agar Cornflour Vegetable oil Dish soap Water	Agar Glycerin PEP Ethylene glycol Water	Agar-gelatin powder Water Safflower oil Propylene glycol Formalin Xanthan gum Liquid detergent	Agar Water Glycerin Silicon Emulsion Sodium Chloride	Agar Glass beads Water Detergent PVP powder Canola oil Coconut oil	Agar Water Oil Surfactant Formaldehyde p-toluic acid l-propanol Alizarin Propylene glycol

Fabrication

Because gel-based phantoms start in liquid form and solidify, moulds are used to form them into their desired shape, with the occasional cutting of the material following the solidification process. Some groups had solid outer shells to contain gel-based materials inside, one using a hemispherical shell [56], while the other used a breast-shaped shell to train ultrasound technicians [102]. Different groups used breast-shaped skin layers [84, 92–94, 99]; four were found to use 3D-printed MRI-derived moulds. However, most groups used hemispherically shaped skin layers to contain inner tissues, with one group also showing a cylindrical phantom [95]. For the articles that commented on the thickness of the skin layer, a common goal of 2 mm was used. When the moulds are pressed together, it should be noted that the process should be done slowly to avoid forming air bubbles.

Several different methods are used for the inner tissue geometry. To represent the heterogeneity of the inner breast tissue, some groups used different mixtures of adipose and fibroglandular material, one combined tissue where the higher the permittivity, the denser the breast [56, 82, 87]. Other groups would pour the adipose mix in either with the pre-shaped fibroglandular components present or by leaving objects, such as straws, in the place of the gland locations. Other groups used hemispherical fibroglandular geometries [75, 89]. Four phantoms contained rod-shaped or conical-shaped fibroglandular components to mimic tissue geometry [85, 96, 97, 100], and three phantoms had "scattered glands" within the adipose material [90, 92, 101], but did not fully explain the geometry.

Three groups used 3D-printed MRI-derived moulds for the breast shape [93, 94, 99], giving the most realistically shaped gel-based phantoms. More importantly, the 3D-printed moulds allow complete tissue geometry knowledge and reproducible geometry. Reproducibility is crucial because it will enable other groups to compare algorithms and recreate the same phantom for future use. However, to allow different groups to use the same phantoms, the *.stl should be open-access. On the other hand, full knowledge of the tissue geometry is not unique to only the MRI-derived phantoms but also the multimodal phantoms since other imaging modalities can be used to image

their tissue geometry.

For tumour geometry, spherical tumours were the most common [56, 94, 99, 100], providing a reasonable geometry to mimic benign tumours. Other tumour phantom geometries include rod or cylindrical tumours [89, 95, 101], hemisphere and semi-oval [75, 81], and cubic [93]. Two of the tumour phantoms were claimed to be cut in any shape [96, 97], so they have the potential to be realistically shaped, as much as a hand-cut tumour can be realistic. In contrast, the most realistically shaped tumour phantom present appeared to be macrolobulated [90], allowing for a malignant-like tumour phantom.

Overall, one main problem with fabricating gel-based phantoms is that they lack versatility. Once the gel-based phantom is made, the inner geometry cannot be changed. In other words, an entirely new phantom must be recreated to alter its density or remove/add a tumour. Furthermore, with increasing interest in algorithmic detection or machine learning, many phantoms must be created to create a large enough dataset. This increases the cost of the gel-based phantoms and adds much additional time to the fabrication process.

2.3.2 Dielectric Properties and Stability

In terms of properties, the gel-based phantoms do relatively well in meeting the desired properties. Table 2.2 shows all publications' average, minimum, and maximum relative permittivity measurements. It should be noted that the values were estimated from any publication that included plots and not tables. Twelve articles compared their TMMs with the target tissue values, demonstrating the ability of gel-based phantoms to mimic the dielectric properties of the breast [81, 83, 87, 89–93, 95, 96, 99].

Regarding dielectric stability, the gel-based phantoms are the least stable compared to solid- or shell-based phantoms. Twelve weeks was the longest for which the phantoms were reported to be stable, and the properties degraded over time [81]. Similarly, the 2005 paper [103] reported that the properties of the phantom were stable for up to nine weeks provided they were not exposed to air.

Table 2.2: Average, minimum, and maximum relative permittivity for the gel-based phantoms presented.

	Skin	Tumour	Fibroglandular	Adipose
Average	37	53	35	7
Minimum	23	41	16	4
Maximum	45	68	55	12

In [87], sunflower oil was used to preserve the phantoms for up to ten days, while [86] used vinegar as a preservative. Although the group in [75] reported that the gel-based phantom was stable for up to eight weeks, they were unsure of the stability, leading them to investigate using Triton X-100 as a more stable liquid-based alternative. Furthermore, in [100], the dielectric properties were reported to be dependent on temperature. Overall, gel-based phantoms require more work to preserve their properties, and the materials should be exposed as little as possible, creating a window of time in which the phantom must be scanned. Because the dielectric properties of gel-based phantoms are unstable over long periods, they increase cost and time since a new phantom will have to be constructed to mimic the breast properly once the lifetime of the material is exceeded.

2.4 Solid-Based Phantoms

Solid-based phantoms are one of the least common phantoms in BMS. The phantoms are either 3D-printed with holes for glandular gel or liquid to fill [104, 105] or carbon-rubber phantoms [77, 106, 107]. For the solid, 3D-printed phantoms, groups have typically moved towards the shell-based design. In contrast, carbon-rubber phantoms are a potential improvement over gel-based phantoms due to their similar properties and extended shelf-life [108].

3D-Printed Solid-Based Phantoms

In [104, 105], 3D-printed phantoms used 3D-printed Acrylonitrile Butadiene Styrene (ABS) to represent the geometry of skin and adipose tissues. The phantoms can accurately represent

the morphology of tissues as they are 3D-printed from MRI-derived *.stl files. The phantoms have the most realistic morphology since they were not modified into shells. However, using ABS as the adipose- and skin-mimicking material results in a lower permittivity material than the respective tissue since ABS has a permittivity of approximately 2.25 [104]. Furthermore, [104] used a 90% Triton X-100 solution to mimic glandular tissue. The solution had a permittivity of 8.1–5.4 over the frequency range of 0.5–3.5 GHz, and the authors claimed that these properties mimicked the contrast between adipose and fibroglandular tissue. Using low-permittivity materials to mimic tissues results in more signals being transmitted into the phantoms, resulting in less realistic experimental scenarios.

In [105], they used a 60% water and 40% gelatine material to represent fibroglandular tissue, which filled the voids in ABS. The fibroglandular material had a higher permittivity (46.17 at 8.5 GHz) than [104]. However, the increased fibroglandular permittivity but the same adipose permittivity result in an unrealistic contrast (46.17:2.714 at 8.5 GHz) between the adipose- and fibroglandular-mimicking materials; thus, unrealistic reflections at the adipose-fibroglandular interface. Water-soaked raisins were used to mimic tumour tissues [105]. The properties were measured at a higher-than-normal frequency range for BMS, with measurements taking place from 6–14 GHz and only a minimal description of how the measurements were made was provided.

Despite having the most realistic morphology and known tissue geometries, using ABS as an adipose-mimicking material limits the experimental utility of the 3D-printed solid phantom. Furthermore, 3D-printing a solid adipose structure limits the tumour placement within the phantom. It strictly limits it to fibroglandular tissue and increases the probability of trapped air bubbles, which can cause experimental uncertainties. The solid design also limits the versatility of the phantom to one geometry.

Carbon-Rubber Phantoms

The fabrication of carbon-rubber phantoms for BMS started after the introduction of the materials as potential TMMs in 2014 [76]. In 2015, the first fabrications of phantoms were introduced [77, 106], with a slight modification of the urethane used introduced by [107], who then focused on the design of tissue-mimicking phantoms. The materials for carbon-rubber phantoms are created by mixing Carbon Black (CB), graphite, and polyurethane, which at high concentrations becomes a dough-like material, then moulding it into the desired shape [76]. However, since polyurethane is a high-viscosity material (approximately 1800 cps), and carbon powders make the mixtures thicker, MRI and Computed Tomography (CT) scans show that the materials have air bubbles inhomogeneously spread throughout the samples [79]. To reduce the viscosity [77], small amounts of acetone can be added to thin the mixtures, but this may weaken the material.

Similar to gel-based materials, carbon-rubber materials can mimic a wide range of properties, similar to the tissues measured from *ex vivo* tissue measurements [76, 77, 106, 107]. Since the skin layer was made to be 2 mm and the skin needs to act as a container, less carbon powder was used to increase the mechanical properties, resulting in the dielectric properties being lower than actual skin [77, 106, 107]. However, it was shown that the carbon-rubber skin layer has reflections similar to gel-based skin layers when placed on adipose-mimicking material at microwave frequencies [108]. In contrast to gel-based materials, carbon-rubber materials were shown to be stable within 5% over eight months [76]. However, the carbon-rubber materials showed a decrease in dielectric properties over two years but insignificant changes in the year following [78].

The skin layer is first moulded to create the phantom, typically a hemispherical shape. The glandular tissue was then moulded and carved into different canonical shapes and placed in the hemispherical skin shell before pouring the adipose mixture into the shell [107]. Leaving placeholders as inserts increase the experimental versatility of the phantoms [77, 106], as the plugs could consist of healthy or unhealthy tissue surrogates. If a liquid is used in place of the adipose mixture, the experimental versatility can be improved [77].

2.5 Shell-Based Phantoms

The design and use of shell-based breast phantoms have been growing in recent years, with the majority of shells being fabricated via 3D-printing or used moulds that were 3D-printed to create the shell [60, 71, 109–118], while one group has used glass [119], and another using a wine glass [120]. Most 3D-printed shells were printed from the MRI breast repository of the University of Wisconsin-Madison [104], allowing the shell-based phantoms to have known tissue geometries and reproducible phantoms. The shells used are then typically filled with liquids that mimic the dielectric properties of the breast, with inner shells used to hold tissue-mimicking liquids that are different from those in the outer shells.

Most of the groups used 3D printed materials for their shell materials [60, 71, 109, 113, 114], while one group used a carbon-silicone mixture [110], and another used a carbon-rubber mixture [111]. As for 3D-printed solid-based phantoms, 3D-printed plastics have low permittivity, so they are not ideal for outer skin shells and have been shown to have significant abnormal responses when used as inner shells [80, 121]. The carbon-silicone mixture used in [110] was shown to have an acceptable permittivity of between 20 and 30 and used conductive 3D-printable materials for the inner shells, but did not consider any polarization effects that can arise from the use of carbon-doped filaments. A skin-mimicking mixture that can accurately represent the dielectric properties of human skin was introduced in [111]. Unfortunately, the authors failed to describe how the mixture was made, saying it was similar to other carbon-rubber mixtures with slight adjustments, and the samples shown have visibly large air bubbles.

Most groups used Triton X-100 solutions or Triton X-100 solutions and glycerin or another liquid for their tissue-mimicking liquids [60, 71, 111–116]. It was shown in [60, 75, 113, 114] that Triton X-100 solutions and glycerin are capable of mimicking the dielectric properties of adipose and fibroglandular tissues, the properties compared to the properties of tissues from *ex vivo* tissue measurements. For tumour phantoms, the material ranged from pork [115, 117] to carbon-resin

[111] to saline-based solutions [60, 71, 112–114, 116]. Overall, the dielectric properties of shell-based materials, i.e., carbon rubber, thermoplastics, and solutions (if properly stored or remade), are stable over time.

Furthermore, using shells allows for a more diverse and extensive dataset with fewer fabrications. In [71], the phantoms consisted of 9 adipose and fibroglandular shells that could be combined to make 66 unique shell combinations, resulting in 1008 different scans when including the rotation of the fibroglandular shell within the adipose shell. The use of liquids within the shells also gives more freedom in where the tumour phantoms may be placed, only limited by the geometry of the shells. However, freedom of tumour placement can result in significant positioning uncertainty.

2.6 Conclusion

Overall, gel-based phantoms use low-cost materials that mimic the breast’s dielectric properties and have been widely used to test BMS systems. However, except for the phantoms that use MRI-derived moulds, the tissue geometry is not well known, and it is difficult to reproduce the same phantom. In addition, they are limited by their short shelf life and lack versatility.

Solid-based phantoms, specifically carbon-rubber phantoms, mimic the dielectric properties of the breast and are more versatile and stable than gel-based phantoms. However, the phantoms typically use simple geometries, and the design can make them more challenging to reproduce and have not as well-known tissue locations as shell-based phantoms.

Finally, shell-based phantoms show the most promise in terms of versatility, allowing the creation of large, diverse datasets. The phantom materials can be considered low-cost, especially since they can be reused, and the liquids and carbon-based shells can mimic the dielectric properties and morphology of breast tissues. However, using 3D-printed thermoplastic poses a problem regarding dielectric properties and unwanted responses from the internal shells.

None of the individual categories of phantoms can produce the ideal phantom. The combina-

tion of materials of different categories may provide an opportunity to create the ideal phantom. For example, using carbon-doped solid materials to replace the dielectric properties of shell materials would improve the skin response of the shells. It could potentially remove the negative response shell from the plastic within the breast. The shell-based design would maintain the complex geometry, easy reproducibility, low cost, and versatile design of the shell-based phantom. In Chapter 3, carbon-doped materials will be investigated to replace the low-permittivity shell materials currently used. In Chapter 4, new liquids capable of mimicking a wide range of properties will be investigated.

3. Investigations into Carbon-Doped Materials for Shell-Based Microwave Breast Phantoms

This chapter aims to investigate carbon-doped materials that could replace the low-permittivity plastics currently used to fabricate the shell of a shell-based breast phantom.

3.1 Introduction

As discussed in chapter 2, shell-based phantoms are advantageous because they are inexpensive, mimic the morphology of the breast, and can create large and diverse datasets. However, some shell materials, 3D-printable thermoplastics, were also observed to have low permittivity and cause unwanted responses [80, 121]. In contrast, others use carbon-doped thermoplastics with higher permittivity [110] but have not considered potential polarization effects, which can even be seen in low-permittivity Polylactic acid (PLA) [122].

On the other hand, carbon-doped resins are capable of mimicking the properties of breast tissue in a solid form [76, 77, 106–108]. These phantoms use the fact that carbon will act as a dielectric when in an insulating matrix [123, 124] - polyurethane in the case of carbon-resin phantoms [76, 77, 106, 107]. However, polyurethane is an elastomer; therefore, compression and stretching of the TMMs can lead to uncertainties in the phantom shape and can be prone to tearing.

As carbon-doped materials are dielectrically accurate TMMs, using those materials as shells in shell-based breast phantoms would be advantageous; specifically, rigid plastic-like materials that

can act as containers for the dielectric liquids used in shell-based phantoms. This chapter will focus on determining the anisotropy of carbon-doped thermoplastics and measuring the dielectric properties of carbon-doped thermoplastics and resins as an alternative to the low-permittivity thermoplastics currently used.

3.2 Methods

3.2.1 Design of a Free-Space Characterization Device

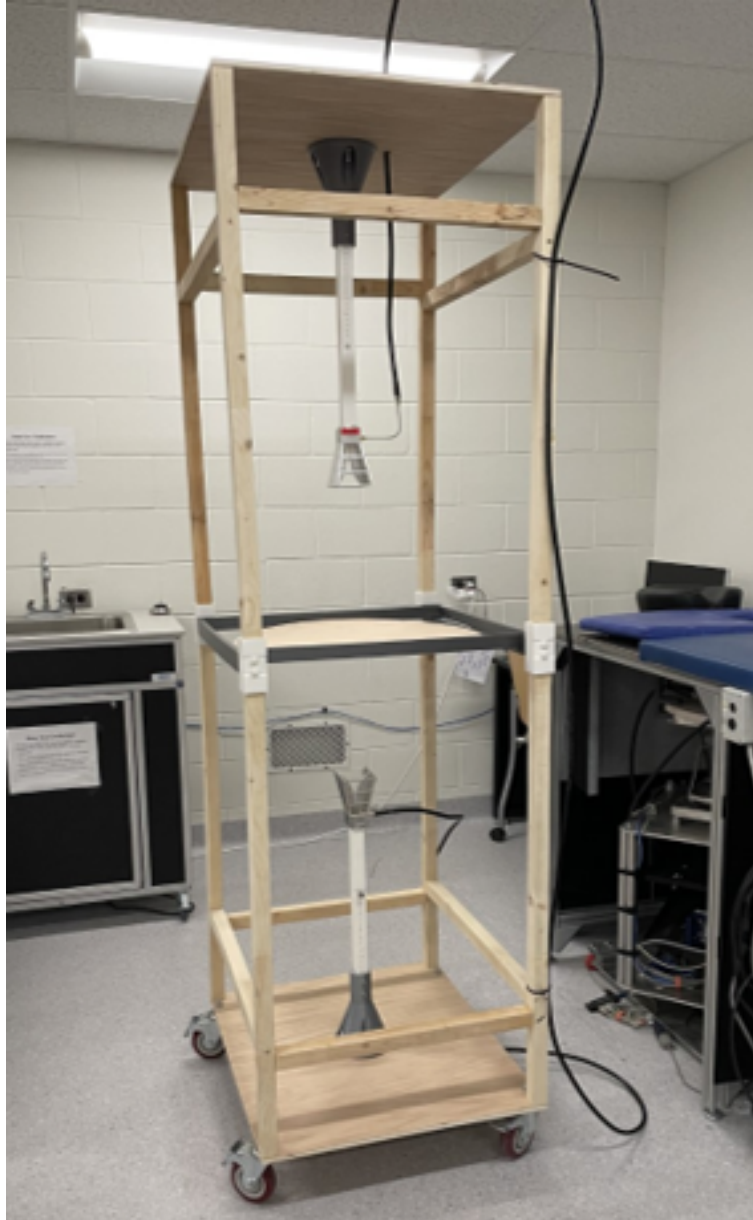


Figure 3.1: The fabricated free-space characterization device, built with plywood bases, a pine 2x2 frame, 3D-printed antenna holder/sliders, and a sample holder. The antennas can be moved in 1 cm increments throughout the 33–65 cm range. The device can hold a 2 mm thick, 60 cm diameter sample centred between the antennas. Only low-permittivity materials are used within the scan region.

A free-space characterization device (illustrated in Figure 3.1) was designed and developed to test 3D-printable materials based on the device in [125]. The device was designed and built to connect to a VNA (C1209, Copper Mountain Technologies, IN, USA) over the 2–9 GHz frequency range. The antennas used were the horn antenna (LB-20200, AINFO, Hong Kong, China), the same equipment used in our pre-clinical bed-styled system [126].

The electromagnetic energy must pass through the Material Under Test (MUT) for dielectric characterization. The device must meet the following requirements for the antenna-to-sample distance or antenna radius, r . This dictates the height requirements of the device [125]:

$$r > \left\{ \begin{array}{l} \lambda, \\ D, \\ FF \text{ (Far Field)}, \end{array} \right\} \quad (3.1)$$

where λ is the wavelength, D is the largest antenna aperture, and FF is the antenna Far-Field (FF), given by

$$FF = \frac{2D^2}{\lambda}. \quad (3.2)$$

Since the antenna radius depends on the wavelength, it will also depend on the frequencies used, meaning that the frequency bandwidth of interest will determine the antenna-to-sample distance. The required antenna to sample distance over the 2–9 GHz range was calculated based on Equation 3.1 and is displayed in Figure 3.2.

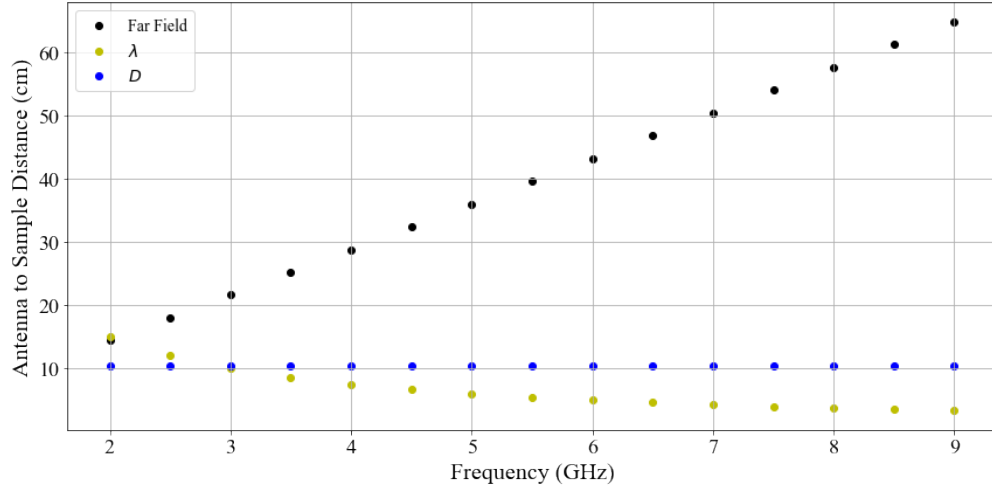


Figure 3.2: The required antenna-to-sample distance as a function of frequency determined by Equation 3.1. The antenna's largest diameter, D , was determined from the antenna specification sheet. The smallest antenna-to-sample distance is 15.0 cm, while the largest is 64.9 cm.

The sample diameter can be obtained from the antenna's 3 dB beamwidth (half-power beamwidth) at the antenna-to-sample distance, as depicted in Figure 3.3. The 3 dB beamwidth depends on the antenna design and is typically given in the antenna's datasheet. The antenna's 3 dB beamwidths were taken from the datasheet, and were used to calculate the required sample diameter, illustrated in Figure 3.4. The scans should be continuous (measured at one radius); however, scanning at the maximum antenna radius, r_{max} would require a sample diameter of ≈ 120 cm, which is an unrealistic sample size. However, if the sample is scanned at $r = 33$ cm from 2–4.5 GHz, then scanned at $r = 65$ cm over 4.5–9 GHz, the sample diameter could be a more reasonable 60 cm. Therefore, the antenna radius, r , was designed to move from 33 cm to 65 cm in 1 cm increments.

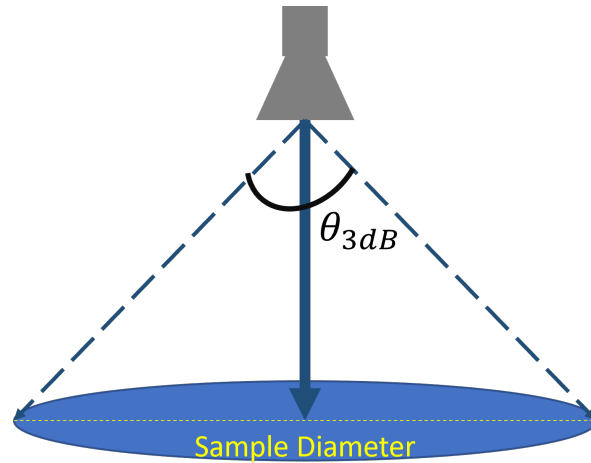


Figure 3.3: The 3 dB beamwidth for the antenna and how the sample diameter must be at least that size. The 3 dB beamwidth was found from the datasheet for the antenna and used alongside the antenna radius to determine the required sample diameter.

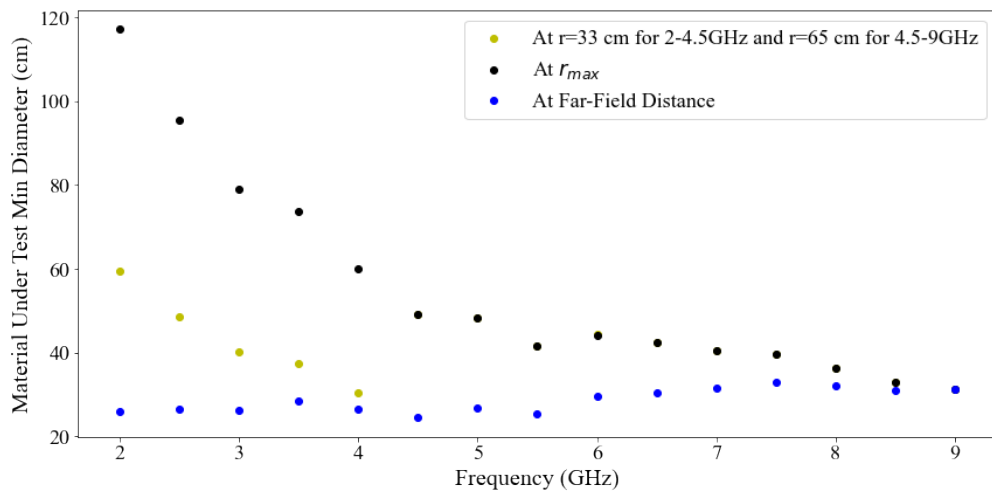


Figure 3.4: The MUT minimum sample diameter as a function of frequency. The blue plot highlights the diameter of the device if it is measured at individual frequencies. The black illustrates the more realistic case, calculating the maximum antenna radius. The yellow line represents the most realistic case concerning the diameter, in which the measurements are made at 33 cm and 65 cm for 2–4.5 GHz and 4.5 –9.5 GHz, respectively.

Since the device works at microwave frequencies, it must be built with low-permittivity materials to ensure minimal reflections from the device itself. Therefore, the following materials were used in the construction of the device:

- Plywood

-
- 2×2 Pine
 - 3D-printed connectors
 - 3D-printed dowels

where all the 3D-printed structures were printed with Polylactic acid (PLA), and the dowels help replace the use of metal screws and bolts within the antenna scan region.

Overall, the device can extract the dielectric properties of 2 mm slabs of solid material over 2–9 GHz. The antennas can be placed at different positions over a range of 33–66 cm; however, attachments can be designed to modify this range. Due to the requirement of large diameter slabs, the device is not ideal for characterizing 3D-printable or expensive materials. The device can still measure smaller materials' transmission and reflection coefficients to characterize how the material will respond to the antenna beam pattern at the frequencies used within a given system. The antenna-to-sample radius of the device can be modified to simulate the response of a material from an experimental BMS scan.

3.2.2 Anisotropy Measurements of Carbon-Doped FDM Materials

Two different types of measurements were taken for carbon-doped Fused Deposition Modelling (FDM) materials: free-space and dielectric probe measurements. The free-space measurements were taken with the device described in the previous section. The materials were 3D-printed in 2 mm thick petri-dish size discs (140 mm in diameter) using less material than the disc size required for dielectric characterization in the previous section.

The 3D-printed materials tested with the free-space characterization device were Conductive Polylactic acid (CPLA), Carbon Fibre Polylactic acid (CF-PLA), and for control, our current 3D-printed shell material, Polycarbonate Co-polyester (PC-CPE). Three total print orientations were used:

1. **Horizontal one-directional print** - The disc is printed flat on the print bed, and the filament is laid in the same direction; this is shown in Figure 3.5

- CPLA and CF-PLA were printed in this orientation to test the polarization of the carbon-doped filament.

2. **Vertical one-directional print** - The disc is printed vertically - in the same direction as the phantom walls would be printed - with the filament laid down in the same direction, shown in Figure 3.6.

- Only the CPLA was printed in this orientation since volume resistivity is different along the x/y-axis (30 ohm-cm) vs. the z-axis (115 ohm-cm) [127], suggesting further dependence on the orientation.

3. **Horizontal cross-hatched pattern** - The disc is printed flat on the print bed, but now the filament is laid perpendicular to the previous layer. Therefore, the orientation will have layers alternating from a 0-degree orientation to a 90-degree; this is shown in Figure 3.7.

- All materials were tested in this orientation.

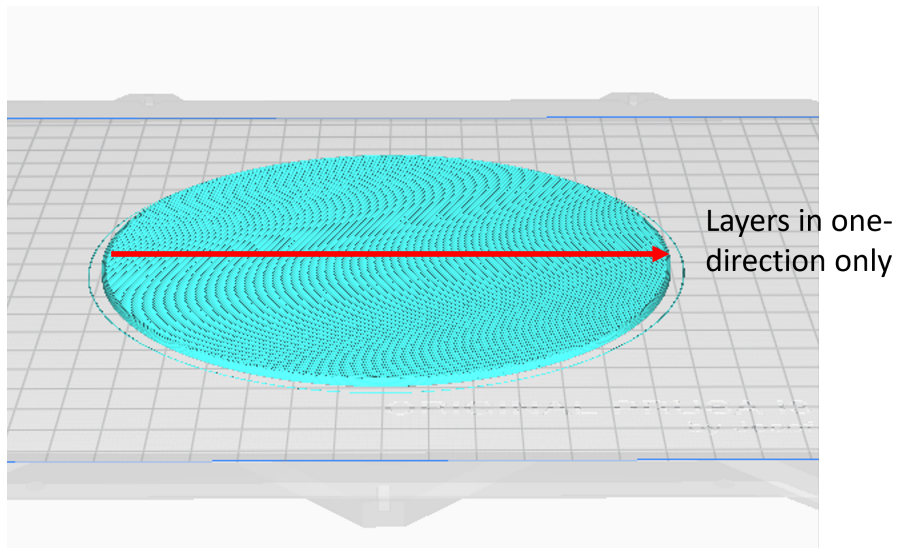


Figure 3.5: The horizontal one-directional print orientation. In this orientation, the printer makes the lines in the same direction.

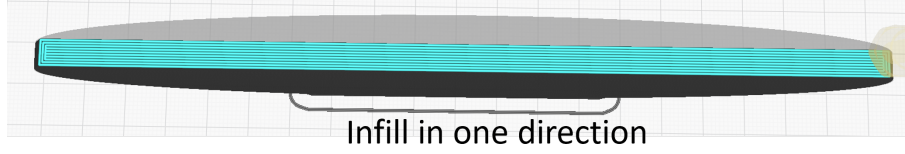


Figure 3.6: The vertical one-directional print orientation. In this orientation, the disc is printed vertically, or how an actual phantom would be printed, giving the most realistic representation of the material responses used experimentally.

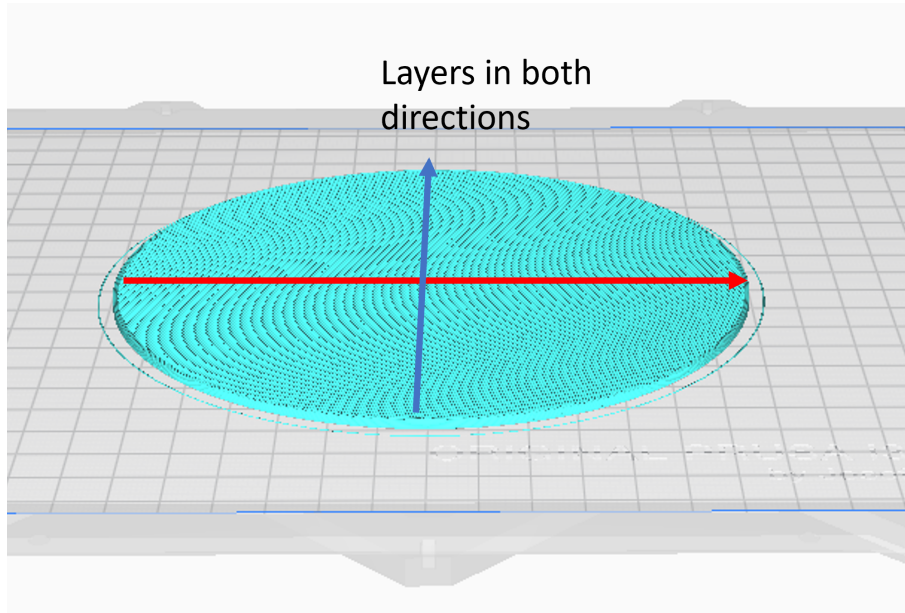


Figure 3.7: The horizontal cross-hatched disc print orientation. In this orientation, the filament lines will alternate directions as they are printed flat on the print bed - the most common flat print infill. It is the equivalent of alternating the horizontal one-directional print after each layer.

The device was first calibrated to the end of the cables using the VNA calibration kit. A three step S -parameter measurement procedure was used to reduce external signals and reflections from the device: air, the MUT, and metal on top of the MUT. The air and metal measurements represent a perfect transmitter and reflector, respectively. The MUT measurement was calibrated using the following equations [125]:

$$S_{11}^{(\text{MUT calib.})} = -\frac{S_{11}^{(\text{MUT})} - S_{11}^{(\text{air})}}{S_{11}^{(\text{metal})} - S_{11}^{(\text{air})}} e^{2j(\omega/c)L}, \quad (3.3)$$

$$S_{21}^{(\text{MUT calib.})} = \frac{S_{21}^{(\text{MUT})} - S_{21}^{(\text{metal})}}{S_{21}^{(\text{air})} - S_{21}^{(\text{metal})}} e^{-j(\omega/c)d}, \quad (3.4)$$

where ω is the angular frequency, c is the speed of light, d is the thickness of the MUT, and L is the thickness of the metal plate. The exponential is to consider the thickness of the MUT and metal plate, while the calibration ensures that the response of the metal is -1 and the air transmission is 0 [125]. Time-domain gating was used in the VNAs software (S2 VNA, Copper Mountain Technologies, IN, USA) to remove any additional responses.

The scan procedure was then:

- 3 air measurements,
- 3 MUT measurements,
- 3 MUT with copper disc measurements.

For CF-PLA and CPLA, measurements were taken every 15° from 0° – 180° to study the anisotropy of the carbon-doped FDM filaments, where 0° was defined as the top layer of filament being in line with the largest aperture of the antenna. Note that PC-CPE was only measured at one angle since any anisotropy would only be detectable by a high-precision method [122]. To analyze the response at each angle, the Area Under the Curve (AUC) of the S -parameter as a function of frequency was determined, given by

$$\text{AUC} = \sum_i^n S_{jk}(i) \quad (3.5)$$

where $S_{jk}(i)$ is the S -parameter at the i^{th} frequency point.

3.2.3 Solids for Shell-Based Breast Phantoms

To determine potential solids for the shell material of shell-based breast phantoms, the properties of carbon-doped FDM filaments and carbon-doped resin were measured with a dielectric probe from 0.4–9 GHz. The FDM filaments were printed in cubes of $5\text{ cm} \times 5\text{ cm} \times 3\text{ cm}$ with the filament laid in one direction (not alternating). The cubes then had measurements made ten times on the top/bottom surface and ten times on each set of parallel sides (denoted side and back) to determine if the properties changed with the print orientation. Figure 3.8 illustrates the top/bottom measurement as perpendicular to the top of the printed cube, the side measurement as perpendicular to the filament from the walls, and the back measurement as parallel with the filament. A PC-CPE sample was also manufactured, and the properties were measured for comparison.

In [76, 77, 106], different mixtures of Carbon Black (CB), graphite powder, and polyurethane were used to create solid phantom materials. However, when mixed, these materials have a high viscosity, which makes them challenging to prepare. The material’s high viscosity allows air bubble development and entrapment during preparation, as can be seen from CT and MRI images of the samples [79]. Therefore, we replaced polyurethane with castable resin (Smooth Cast 327) because the resin has a lower viscosity (300 cps [128] compared to 1800 cps for polyurethane [129]). The exact material and provider are in Table 3.1. In addition, the resin cures into a Shore D hardness material, improving the mechanical structure of TMMs, which is necessary to hold the tissue-simulating liquids and maintain a controlled shape.

Table 3.1: Suppliers for carbon-doped resin materials.

Material	Supplier	Location
Carbon Black, 50% compressed	Fisher Scientific	Hampton, NH, USA
Graphite, powder, $<20\text{ }\mu\text{m}$	Sigma-Aldrich	St. Louis, MO, USA
Resin, Smooth-Cast 327	Smooth-On	Easton, PA, USA

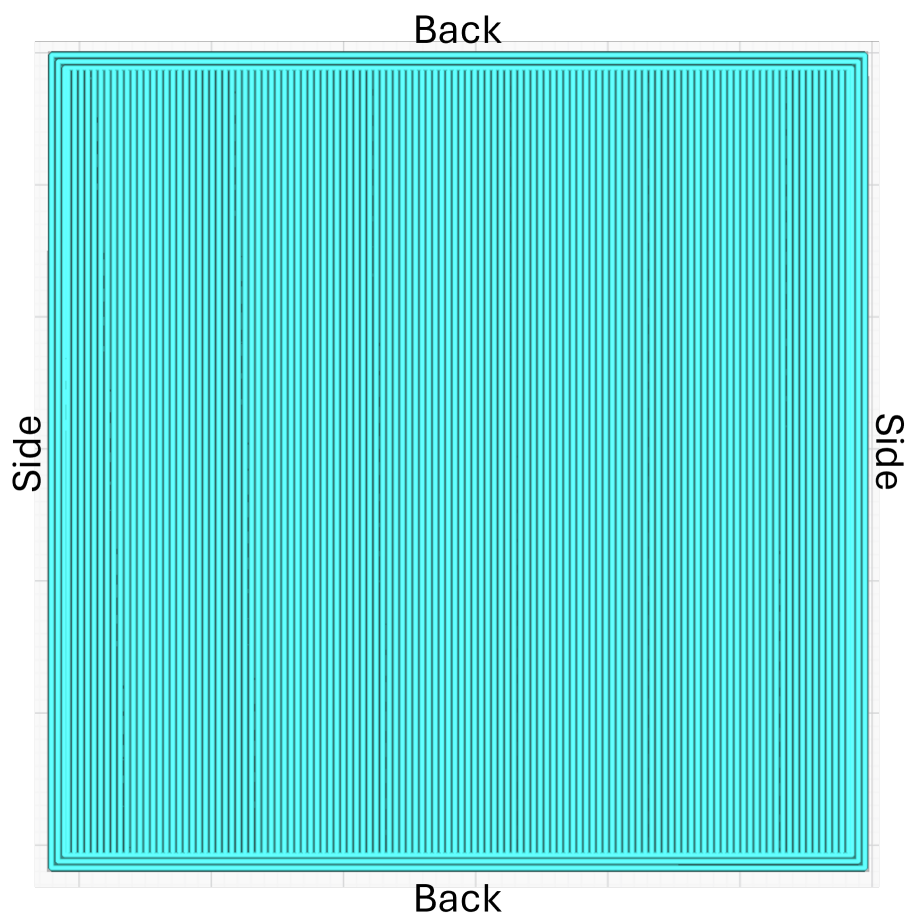


Figure 3.8: View from the top/bottom of the printed cube illustrating that the side measurements are perpendicular to the filament, while the back measurement is taken parallel to the filament

Table 3.2: Concentrations of the materials used in the carbon-doped resin materials (% weight). The two mixtures for adipose-mimicking materials are to be compared.

Tissue	% Weight CB	% Weight Graphite	% Weight Resin
Adipose	3%	0%	97%
	0%	15%	85%
Skin	3%	35%	62%

A solid skin-mimicking mixture and two solid adipose-mimicking mixtures were prepared in a fume hood; concentrations can be seen in Table 3.2. No fibroglandular-mimicking mixture was produced in this manner the high concentrations of carbon required would affect the mechanical stability of the shell. Theoretically, an adipose-mimicking material would work in constructing the shell since the tissue-mimicking liquid and solid would have matched or close to matching properties, minimizing the response due to a reduced property mismatch. All mixtures were pressed into a 5 cm diameter $\times$ 12 cm high cylindrical mould. Once cured, the sample was cut into three parts using a lathe and wet sanded to obtain a smooth, flat surface. Ten measurements were performed on each surface, with the probe-MUT interface being inspected for air gaps, giving 60 total measurements and allowing the homogeneity of the samples to be determined.

All mixtures were created using the following procedure:

1. Weigh all the components.
2. Add graphite to the mixing bucket.
3. Combine Part A and Part B of the resin. Mix until it is homogeneous.
4. Pour the resin into the mixing bucket on top of the graphite. This will trap the graphite beneath the resin and speed up mixing. Mix until the mixture appears homogeneous.
5. Slowly add the carbon black while stirring the mixture – this will help homogeneously mix the carbon black into the resin.

The adipose-mimicking mixtures were mixed using metal stirring rods. For high concentrations of carbon powder, i.e., approximately 8% CB, 32% graphite, or 30% graphite and 3% CB, the mixture became very thick, like a dough, and had to be mixed by hand using chemical-resistant gloves until it was visually homogeneous. In contrast, the adipose-mimicking mixtures remained in a liquid form.

3.2.4 Characterization of Dielectric Properties

To characterize the dielectric properties of potential TMMs, an open-ended coaxial dielectric probe was used (DAK 3.5, SPEAG, Zurich, Switzerland), along with the same VNA (C1209, Copper Mountain Technologies, IN, USA) used in the pre-clinical bed system. All measurements were made over the frequency range of 0.4–9 GHz, which covers the frequencies used by the system in [126] experimentally. A dielectric probe was used over the resonator method since it allows measurement across more frequency points.

An open-short-load (OSL) calibration procedure was used to calibrate the dielectric probe. The open measurement was in air, while the short used a thin copper strip provided by SPEAG. The load measurement was performed with deionized water at a measured temperature. To verify calibration, a calibration liquid with known properties was used (HBBL600-10000V6, SPEAG, Zurich, Switzerland). The calibration was deemed successful if the measured properties of the calibration liquids were within $\pm 3\%$ of their known properties.

A screw-based mechanical lift stage created contact between the probe and the MUT. For solid materials, the contact between the probe and the material was inspected for any gaps (i.e., light coming from between the contact) to ensure good contact between the probe and MUT. The measured properties of the potential solid TMMs were compared to the modelled properties of *ex vivo* tissue measurements in [30, 130] to determine their experimental validity. The dielectric measurements' uncertainty was the maximum of the probe's inherent uncertainty (3%) or one standard deviation from the measurements. Significant deviations were observed for the carbon resin mate-

rials due to inhomogeneities caused by tiny air bubbles and the distribution of the powders within the resin.

3.3 Results and Discussion

3.3.1 Anisotropy Measurements of Carbon-Doped FDM Materials

The free-space characterization device was used to characterize the anisotropy of carbon-doped FDM materials. The minimum antenna radius of 33 cm was used to obtain the largest response for smaller sample sizes. The S -parameter Area Under the Curve (AUC) results for the two orientations of CF-PLA is illustrated in Figure 3.9, the results for the three orientations of CPLA are illustrated in Figure 3.10. Table 3.3 gives the maximum percent difference in the S -parameter AUC over all angles, quantifying the anisotropy.

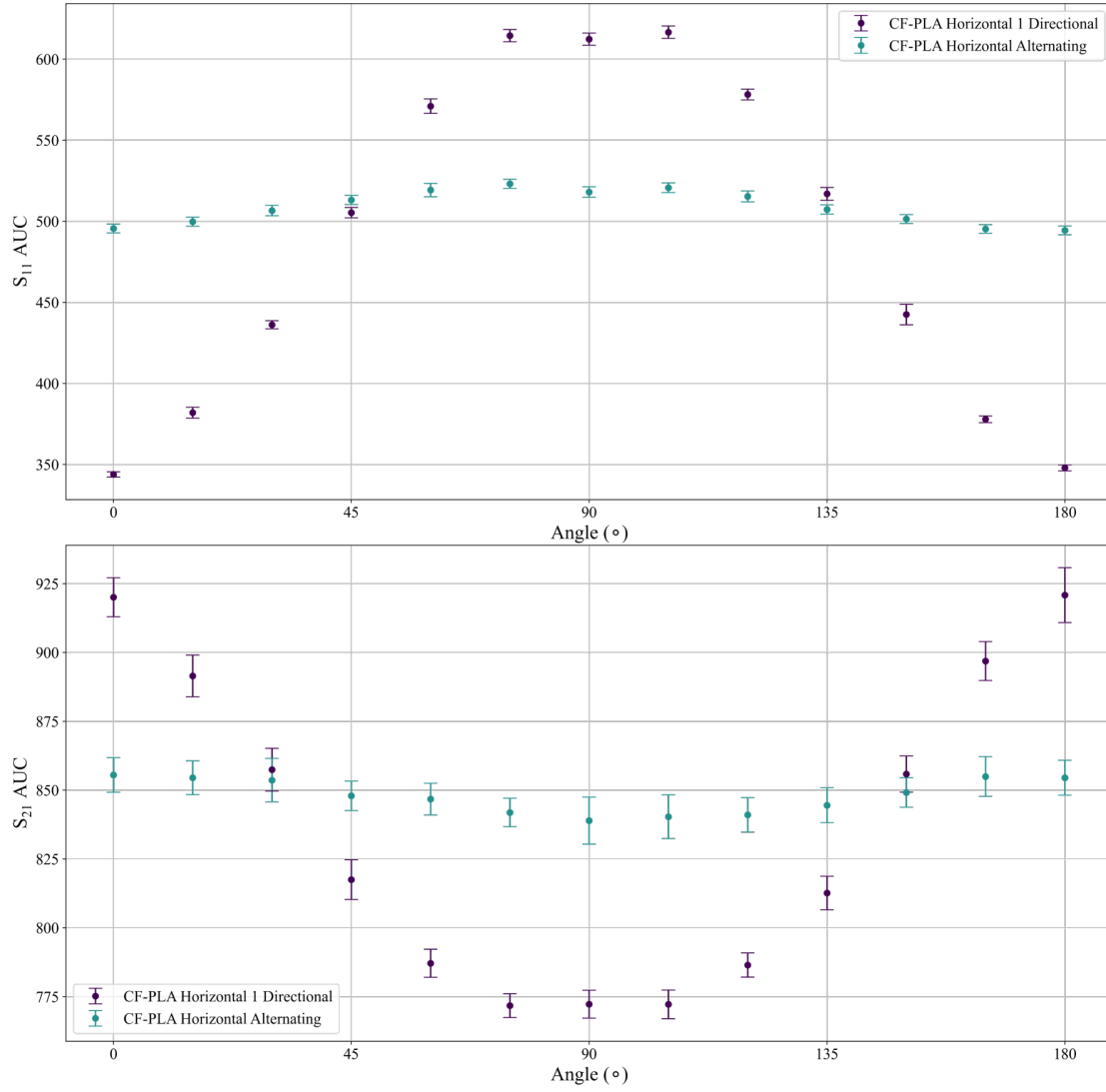


Figure 3.9: The S_{11} and S_{21} AUC as a function of antenna angle with respect to the MUT from 0–180° for CF-PLA. The polarization is significant when the filament is aligned in the same direction as the largest antenna aperture.

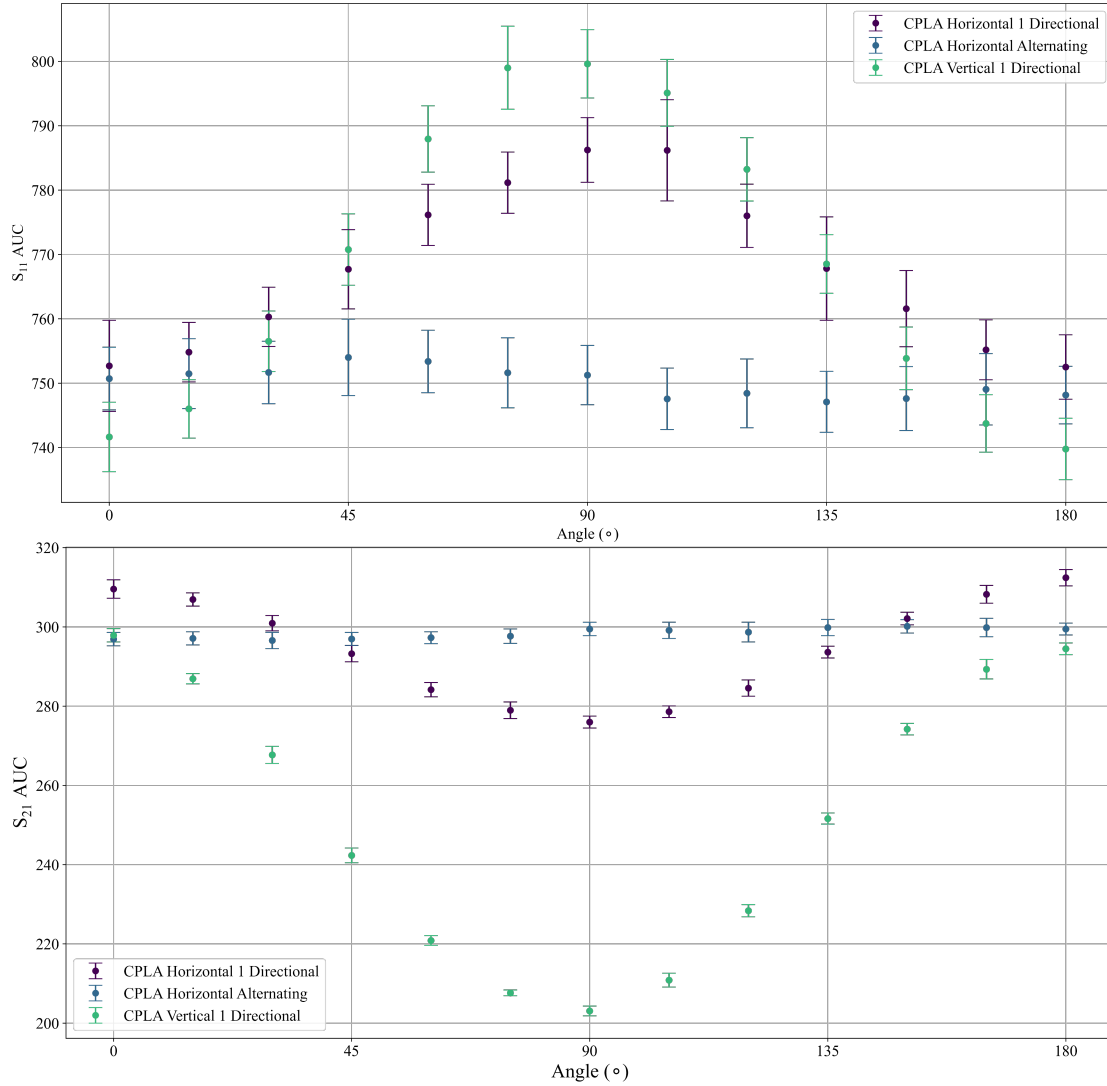


Figure 3.10: The S_{11} and S_{21} AUC as a function of antenna angle with respect to the MUT from 0–180° for CPLA. The figure illustrates the polarization whenever the filament is aligned in the same direction. Still, the effects are most prominent when the sample is printed vertically, displaying the differences in the volume resistivity from the technical specifications [127].

Table 3.3: The maximum percent difference of the S-parameter AUC for each material in each retention, quantifying the anisotropy of the MUT.

Material	Print Orientation	Max. % Change in S_{11} AUC	Max. % Change in S_{21} AUC
CF-PLA	Horizontal 1-Directional	80(2)%	19(2)%
	Horizontal Cross-Hatched	5(1)%	2(2)%
CPLA	Horizontal 1-Directional	4(1)%	13(2)%
	Horizontal Cross-Hatched	1(1)%	1(2)%
	Vertical 1-Directional	8(1)%	47(3)%

From Table 3.3 and Figure 3.9, it can be seen that as the disc is rotated between 0° and 180° , there is significant polarization when the CF-PLA is aligned with the antenna's polarization. On the other hand, when the CF-PLA is printed with layers alternating direction (0° to 90°), the alternating directional layers will result in more constant AUC (3/4 maximum percent changes agree with 0%) with respect to the angle since the polarization effects of each layer will average out.

For CPLA, there is only a 4(1)% change in S_{11} -AUC between 0 – 90° for a horizontal directional print whereas, for the CF-PLA, the change is 80(2)%. Figure 3.10 illustrates that for the S_{11} measurements, the change is more significant than the uncertainty in the measurements, suggesting that polarization is still present and will affect the measurements. However, the measured S_{21} for the vertically printed CPLA disc (how the phantom would be printed) has a significant change in signal (47(3)%), indicating that phantoms printed with this material would experience polarization effects. It should also be noted that there is a difference in the horizontal and vertical print orientations, suggesting that different properties depend on how the material is printed. Similar to CF-PLA, the cross-hatched print shows a more consistent AUC due to the layers alternating from 0 to 90 degrees.

These results show that carbon-doped FDM filaments are polarized at microwave frequencies. The orientation of the antenna and how the material is printed impact the material's response at microwave frequencies. When used in the fabrication of phantoms, the response will be less controlled and may vary from system to system as the morphology of the phantom changes, i.e., the angle at which the microwaves will hit the phantom walls. From this, it can be said that the dependence of the carbon-doped filaments on their orientation to the antenna makes them unsuitable for BMS phantoms.

3.3.2 Solids for Shell-Based Breast Phantoms

Figure 3.11 displays the dielectric properties of the 3D-printable materials with measurements taken from the top (perpendicular to the filament from the top and bottom surfaces), side (perpen-

dicular to the filament from the wall), and back (parallel to the filament from the wall). Figure 3.12 displays the dielectric properties of the carbon-doped resin materials. Both plots contain the properties of carbon-rubber materials from [106] (referred to as "reference"), and the mean properties of *ex vivo* measurements from [130] and [30] for adipose and skin, respectively, referred to as the "target". The Mean Absolute Percent Error (MAPE) between the TMMs and modelled tissue properties was calculated. Table 3.4 displays the MAPE between tissue and carbon resin TMMs over the 0.4–9 GHz frequency range.

Table 3.4: Mean absolute percent error between tissue and TMMs

Adipose		
Material	ϵ'	σ
PC-CPE	69(2)%	99.49(5)%
15% Graphite - 85% Resin	20(4)%	88(2)%
3% CB - 97% Resin	30(20)%	50(20)%

Skin		
Material	ϵ'	σ
PC-CPE	93.0(4)%	99.84(2)%
35% Graphite - 3% CB - 62% Resin	30(10)%	30(30)%

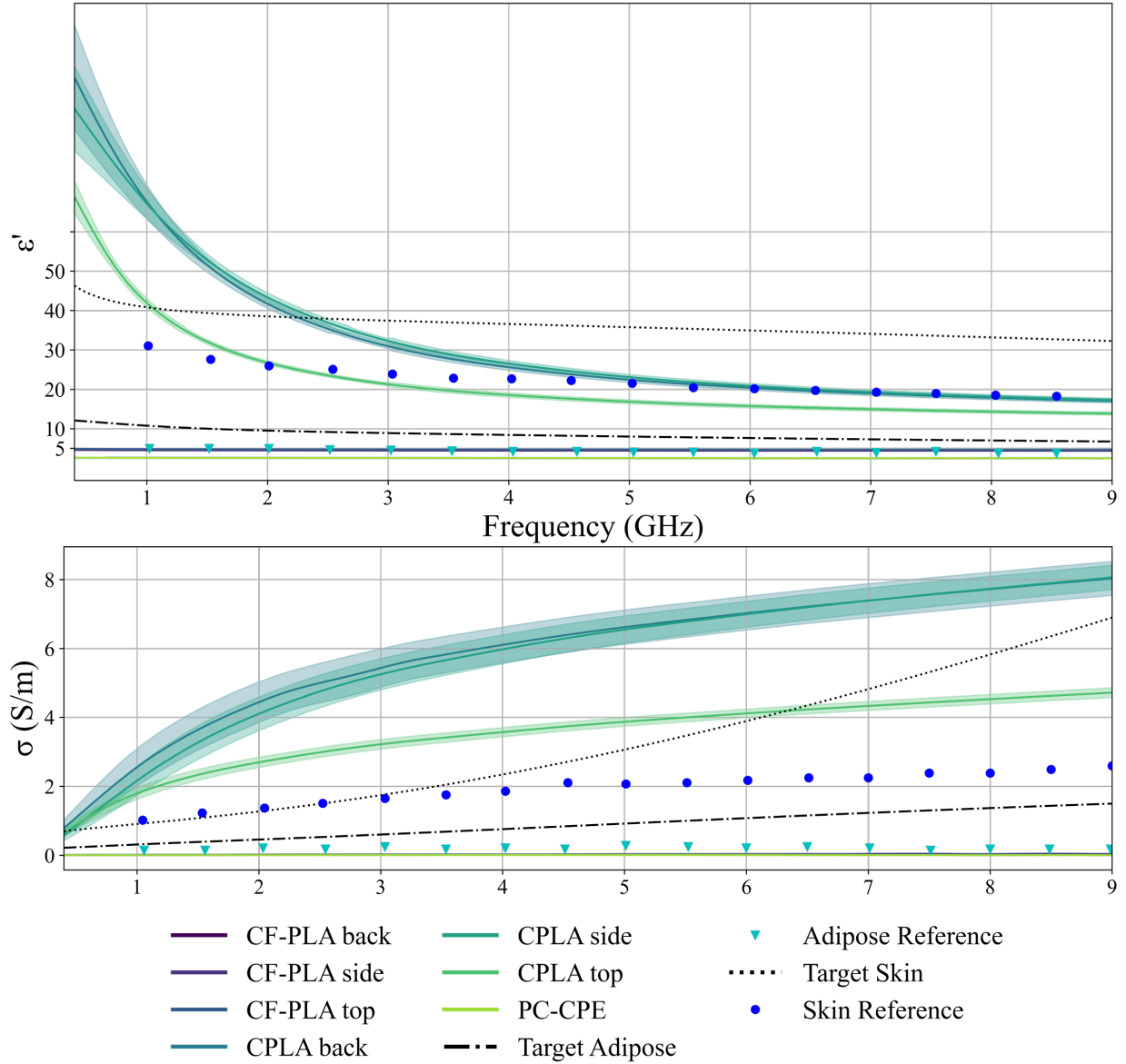


Figure 3.11: The dielectric properties of the 3D-printed CF-PLA and CPLA measured from the top (perpendicular to the filament from the top and bottom surfaces), side (perpendicular to the filament from the wall), and back (parallel to the filament from the wall). All of the CF-PLA measurements were in agreement with one another, indicating there is no dependence in the properties on the print orientation. It can be seen that the properties of the CPLA disagree when measuring from the top/bottom (the surface on the print bed) and the vertical walls (side and back), indicating an orientation dependence of the dielectric properties.

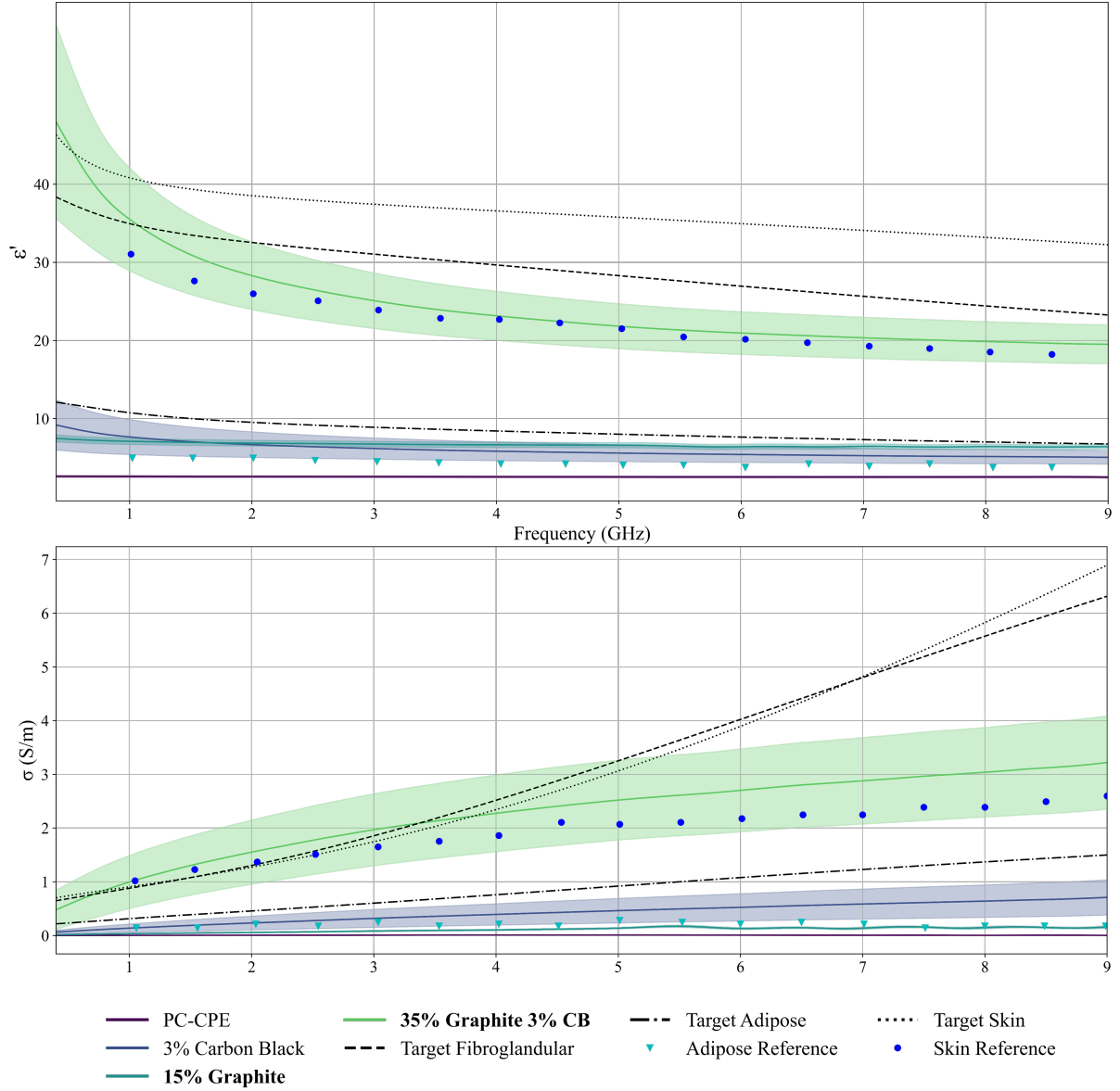


Figure 3.12: The dielectric properties of the carbon-doped resin created in this work. For the adipose-mimicking material, the dielectric properties are an improvement over those of the material from [106] (adipose reference), while the skin-mimicking materials (skin reference) have similar dielectric properties.

Figure 3.11 shows that all the CF-PLA measurements agree across each orientation, illustrating that the dielectric properties are not dependent on the print orientation. Conversely, the CPLA measurements depend on the print orientation, as the side/back walls were found to have different properties than the top/bottom surface of the sample. Based on the anisotropy measurements in the

previous section and the changes in the CPLA for different orientations, it can be concluded that these materials are not suitable as shell-based phantom materials.

For skin, carbon resin reduces the MAPE of the permittivity from 93.0(4)% to 30(10)% and the MAPE of the conductivity from 99.84(2)% to 30(30)% when compared to PC-CPE. The properties of the carbon-resin TMMs agree with the carbon-polyurethane skin-mimicking material properties from [106]. For the adipose layer, both materials reduced the MAPE compared to the PC-CPE values of 69(2)% and 99.49(2)% for the permittivity and conductivity, respectively. The MAPE became 20(4)% and 88(2)% for the 15% graphite-resin and 30(20)% and 50(20)% for the 3% CB-resin, the permittivity and conductivity, respectively.

Figure 3.13 displays 60 individual measurements taken from various points of both potential adipose materials'. The 3% CB mixture has considerable variation in ϵ' compared to the graphite-resin, suggesting that the graphite-resin is more homogenous than the CB-resin. When mixing with resin, clumps of the CB were found to form. Figure 3.13 suggests that these areas with high concentrations can cause spikes in permittivity. In contrast, graphite did not have any clumping when mixed with the resin, which is reflected by less variability in ϵ' for Graphite in Figure 3.13.

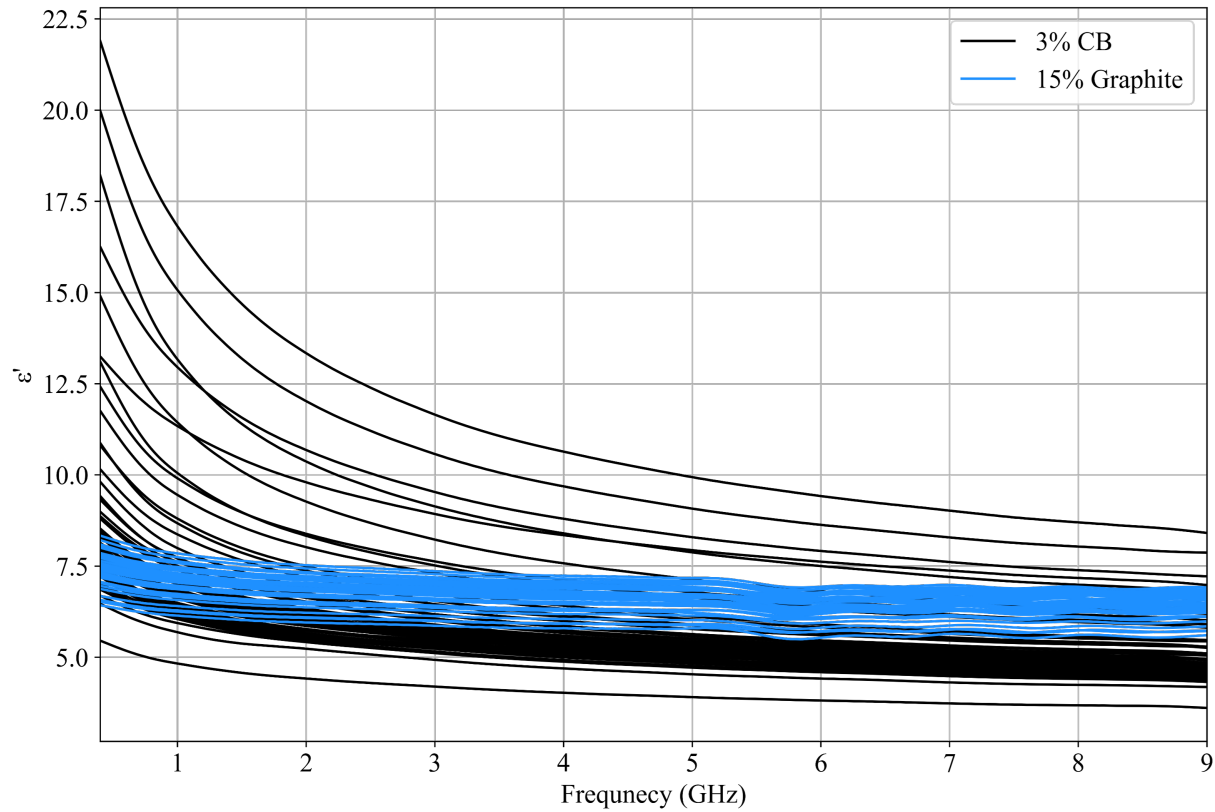


Figure 3.13: The 60 individual permittivity measurements of 3% CB and 15% graphite resin mixtures. The permittivity of the 3% CB mixtures varies widely throughout the three samples, while the 15% graphite mixtures have less variation.

In general, resin-based TMMs have improved durability and mechanical properties necessary to act as a liquid container, which polyurethane-based lack. However, because of the complex geometry of the fibroglandular shells, the adipose-mimicking material could not be cast into the shapes of those shells. The shell geometry would have to be more hemispherical to allow moulding. Therefore, other methods must be investigated to create dielectrically accurate inner shells that allow for complex geometries. Future work needs to be done to mould the skin-mimicking materials into the shapes of the skin shells given in [71].

3.4 Conclusion

A free-space characterization device was designed and fabricated for microwave frequencies. The device can measure the S -parameters and extract the dielectric properties of different MUTs. The fabricated device was developed with low-permittivity materials to reduce the responses from the device. It can be used to test the response of materials at different antenna-to-sample distances and can also be modified to use other antennas used in the laboratory.

The free-space characterization device was used to scan 3D-printable, carbon-doped FDM filaments. From the results, the carbon-doped filaments were shown to have anisotropy at microwave frequencies. For CPLA, which was used as a phantom material in [110], the percent change in the S_{11} and S_{21} measurements was 8(1)% and 47(3)%, respectively. The results mean that the response will depend on the angle of the antenna, which will change in a breast phantom from the chest wall to the nipple, making it not a well-controlled material. For that reason, it is recommended that carbon-doped filaments are not used for the fabrication of phantoms in BMS.

Carbon-doped resin was also studied as a potential shell material. The resin improved the MAPE of the properties of the current skin shell material from 93.0(4)% to 30(10)% and 99.84(2)% to 30(30)% in terms of permittivity and conductivity, respectively. Similarly, the inner shell properties could be improved from 69(2)% to 20(4)% and 99.49(5)% to 88(2)% for the permittivity and conductivity compared to adipose tissue. Using resin as a base had comparable properties to that of [106], which used polyurethane. Using resin gives the material a lower viscosity, reducing the likelihood of air bubbles. It cures into a more rigid material, creating a better container for liquids when compared to polyurethane. Overall, it gives dielectric properties comparable to those of carbon-rubber phantoms but has more desirable mechanical properties to be used as a shell in shell-based phantoms.

Carbon resin can be cured into skin shells by 3D-printing moulds of the skin shells and pressing the material between them until it cures. However, future work must determine how the inner

shell, which has a complex geometry, can be fabricated. Potentially, moulds could be printed with small supports between them; the supports could then be drilled out and filled with the same resin mixture. Further studies would then need to be conducted to:

1. Determine the response of the skin material when overlayed on adipose tissue, as in [108].
The free-space characterization device could be used in this study to get the S -parameter measurement with different antennas and materials and compare it with the actual response of skin.
2. Determine the impact of using an adipose-like dielectric shell for the inner phantom shell in an experiment similar to [80, 121].

In summary, this chapter had the following outcomes:

1. A free-space characterization device was developed that can measure the S -parameters of different materials.
2. Carbon-doped FDM filaments were shown to have undesirable anisotropy at microwave frequencies, making them unsuitable for BMS phantoms.
3. 35% Graphite - 3% CB - 62% Resin was determined to be a suitable skin-mimicking material that could be pressed into a skin shell for BMS.
4. 15% Graphite - 85% Resin was determined to be a suitable adipose-mimicking material, but future work must be completed to determine how the material could be manufactured into inner breast phantom shells.

4. Investigations into New Tissue-Mimicking Liquids for Breast Microwave Sensing

This chapter investigates new, non-viscous tissue-mimicking liquids that can be used in breast microwave sensing.

4.1 Introduction

Glycerin and a 30% Triton X-100 solution have been widely used to mimic adipose and fibroglandular tissue. Unfortunately, these liquids have a high viscosity, which increases the probability that air bubbles will be present during a scan, especially when the liquids get agitated, i.e., when moving shells or tumour phantoms. Due to its water-like viscosity, Diethylene Glycol Butyl Ether (DGBE) solutions of 0–100% concentration with water were investigated as potential TMMs.

4.2 Methods

4.2.1 Liquids for Shell-Based Breast Phantoms

DGBE solutions were investigated as a replacement for glycerin and Triton X-100, currently used as adipose and fibroglandular-mimicking liquids. Liquid samples were prepared with varying percent-by-weight concentrations of DGBE and water ranging from 0–100%, in increments of

10%. For comparison, samples of the liquid glycerin and Triton X-100 were made. The properties of the solutions were then measured and compared to modelled properties from *ex vivo* tissue measurements [130]. DGBE solutions were prepared by pouring them into the water and stirring until visually homogeneous.

In contrast, the Triton solution was prepared by combining it with water and using a magnetic mixer on a hot plate at approximately 30°C for 1 hour. After the solution was homogeneous, it was cooled in a fridge to rest. Before measurements were taken, the solution was left in the air until it was at room temperature.

4.2.2 Characterization and Modeling of Dielectric Properties

Dielectric measurements were performed using an open-ended coaxial probe (DAK 3.5, SPEAG, Zurich, Switzerland) and a vector network analyzer (C1209, Copper Mountain Technologies, IN, USA) of 0.4 to 9.0 GHz. Due to their well-known properties, the probe was calibrated using an open-short-load calibration method using air, a thin copper strip, and water. A verification liquid (HBBL600-10000V6, SPEAG, Zurich, Switzerland) with known dielectric properties was used to confirm calibration. The calibration was said to be successful if the measured dielectric properties were within $\pm 3\%$ of the known dielectric properties of the verification liquid.

The liquid-filled containers were placed on a mechanical lift stage and raised until the head of the dielectric probe was completely submerged. The probe head was inspected for air bubbles, and any were removed using a cotton swab. Ten measurements were taken for each liquid. The dielectric measurements of the liquids were observed to have a lower standard deviation than the probe's inherent uncertainty of $\pm 3\%$.

The measured properties of the liquids were compared to the modelled average properties of *ex vivo* adipose and fibroglandular tissue measurements reported in [130]. However, it should be noted that dielectric properties have been shown to vary widely between individuals [30, 32–34]. Therefore, TMMs should also be able to mimic a wide range of values, allowing the experimental

results to better mimic what may be seen clinically.

For the DGBE solutions, a one-pole Cole-Cole equation was used over the 0.4–9 GHz range to model the complex relative permittivity from 0–100%, ϵ_r ,

$$\epsilon_r = \epsilon_h + \frac{\epsilon_s - \epsilon_h}{1 + (j\omega\tau)^{1-\alpha}} \quad (4.1)$$

where ϵ_s and ϵ_h are the dielectric constants at low (static) and high (infinite) frequencies, ω is the angular frequency, τ is the relaxation time, and α is an empirical factor, where $0 \leq \alpha \leq 1$. The parameters for each DGBE solution were found at concentrations of 0–100%, increasing by 10% for each measurement. Mathematical models of the parameters were created as a function of DGBE concentration, as done in [131] with Ethylene Glycol. To find the optimal Cole-Cole parameters, the cost function, F , was minimized, where F is given by

$$F = \sum |c_\epsilon - m_\epsilon|, \quad (4.2)$$

where c_ϵ and m_ϵ are the calculated and measured complex permittivity at each frequency, respectively.

4.2.3 Stability and Restoration Measurements

Temporal measurements of the glycerin, Triton solution, and 35% and 95% DGBE solutions were made to determine the temporal stability of the solutions. The two latter solutions were used as they have properties similar to those of the former, giving a robust comparison. The solutions were prepared and stored in two containers, one open and one closed. Measurements were made weekly over twelve weeks. In the first weeks, it was observed that all liquids in open containers except glycerin were experiencing physical changes. The Triton X–100 solution became gel-like, and the DGBE solutions experienced significant decreases in volume due to what is hypothesized to be evaporation. Therefore, solutions stored in open containers were excluded from further analysis.

The thermal stability of the 35% and 95% DGBE solutions was analyzed to determine how well the properties are known at room temperature, the temperature at which the solutions are intended to be used. The solutions were refrigerated to approximately 12°C and then heated on a hot plate to increase the temperature to roughly 30° C. The temperature was measured with a temperature probe immediately before dielectric measurements. Since the solutions are intended to be used around room temperature (21.9°C), the properties were compared to those at room temperature $\pm 3\%$, the uncertainty of the probe.

Since solutions may experience evaporation, the DGBE solutions were left out over three days to determine how much evaporation occurred. The properties were measured before and after those three days to assess the effects of evaporation. If the properties of the solutions changed, water was added to the solution in small increments to determine if the properties could be restored to their original value, i.e., to see if the solutions could be restored after use and thus could be used on multiple occasions.

4.3 Results and Discussion

4.3.1 Dielectric Measurements of DGBE Solutions

Figure 4.1 displays the dielectric properties of DGBE with varying concentrations from 0–100%. The Cole-Cole parameters from Equation 4.1 were extracted for each concentration and are shown in Table 4.1. A one-pole Cole-Cole model was used due to the non-zero α parameter observed for the DGBE solutions, suggesting that the one-pole model was required. The one-pole model resulted in the MAPE between the modelled and measured properties not exceeding 1.5%; therefore, it accurately modelled the solutions' properties.

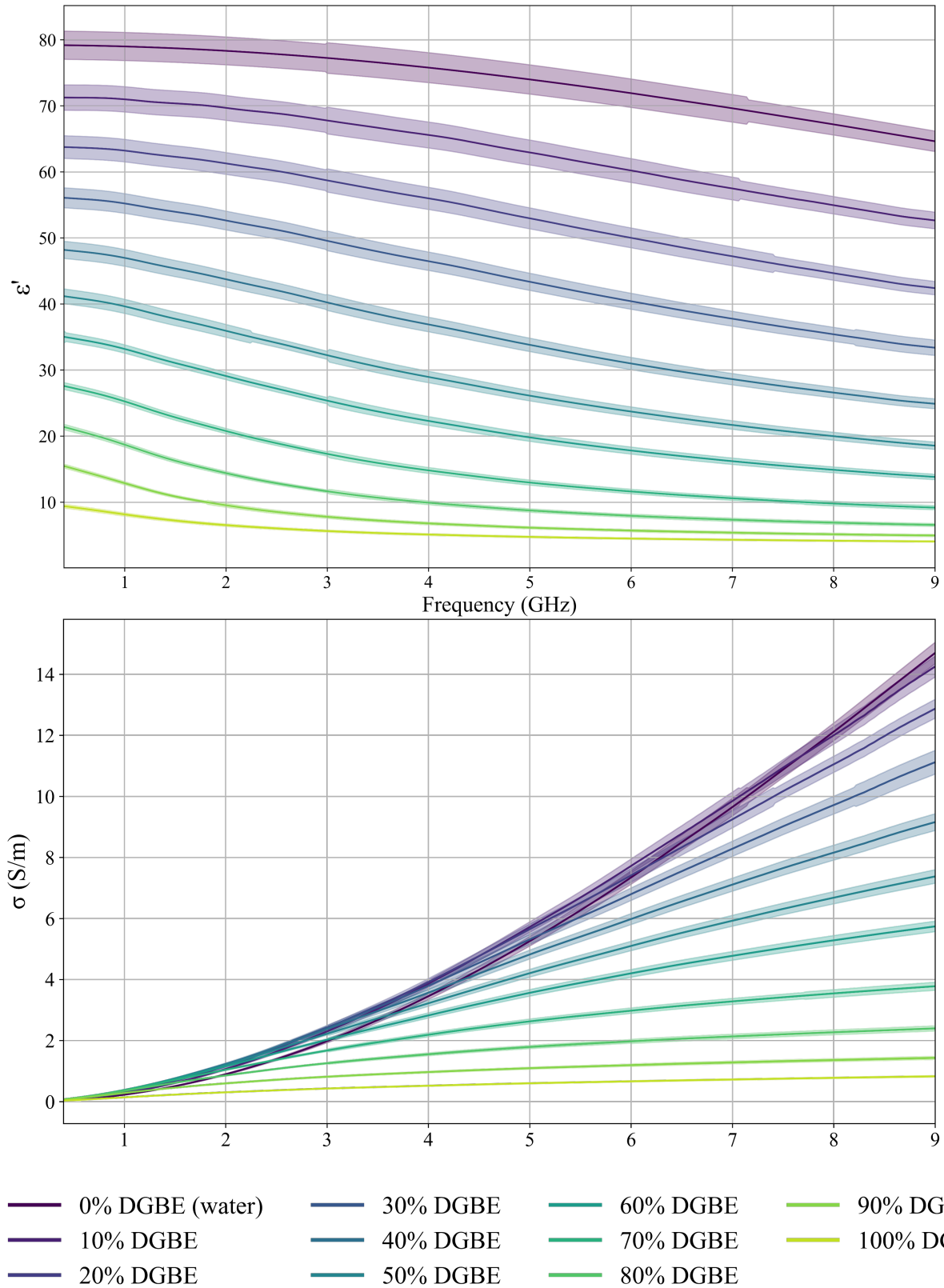


Figure 4.1: The dielectric properties of DGBE solutions from 0 to 100% show the wide range of properties the solution can mimic.

Table 4.1: Optimal Cole-Cole parameters for each solution of DGBE

% DGBE	ϵ_h	ϵ_s	$\tau[ps]$	α
0%	5.45	79.20	8.78	0.00
10%	3.52	71.80	10.68	0.05
20%	3.72	64.57	12.87	0.08
30%	4.03	57.14	15.76	0.11
40%	4.77	49.42	20.73	0.13
50%	4.33	42.60	26.00	0.14
60%	4.34	36.58	33.60	0.14
70%	4.17	29.37	49.68	0.15
80%	3.85	23.51	71.64	0.15
90%	3.40	17.93	101.75	0.18
100%	2.96	10.73	91.14	0.22

The properties were then mathematically modelled as a function of DGBE concentration from 10–100% due to the change in the trends of the parameters at low concentrations. The models for each parameter are given in Equation 4.3–Equation 4.6, where x is the concentration from 0.1 to 1. A piecewise expression was required for ϵ_h , and two linear equations were used: one for concentrations from 10–53% and the other for concentrations from 53–100%. Similarly, a linear equation modelled ϵ_s and τ , while α was modelled with polynomials. These models were as follows:

$$\epsilon_h = \begin{cases} 2.66x + 3.28, & x \leq 0.53 \\ -3.54x + 6.58, & x > 0.53 \end{cases} \quad (4.3)$$

$$\epsilon_s = -68.13x + 77.96 \quad (4.4)$$

$$\tau = 110.4x^2 - 16.5x + 9.8 \quad (4.5)$$

$$\alpha = \frac{1}{100} (62.4x^3 - 104.6x^2 + 64.3x - 0.3) \quad (4.6)$$

The Cole-Cole parameters were then predicted for 35%, 50%, and 95% DGBE solutions, and the properties were determined using Equation 4.1. The modelled properties of Equation 4.3–Equation 4.6 used in Equation 4.1 were compared with the measured properties of each solution

in Figure 4.2, illustrating the ability to determine the theoretical properties of a DGBE solution for a given concentration. The MAPE of the permittivity and conductivity between the measured and modelled Cole-Cole properties were then determined and displayed in Figure 4.3. The predicted permittivity and conductivity uncertainty were determined since the coefficients in Equation 4.3- Equation 4.6 were derived using curve fitting. The relative uncertainty in the predicted conductivity was highest for low concentrations of DGBE, which explains the trend observed in Figure 4.3, while the higher uncertainty in the conductivity of 100% DGBE's is due to higher uncertainties in the conductivity of the original fit from Table 4.1.

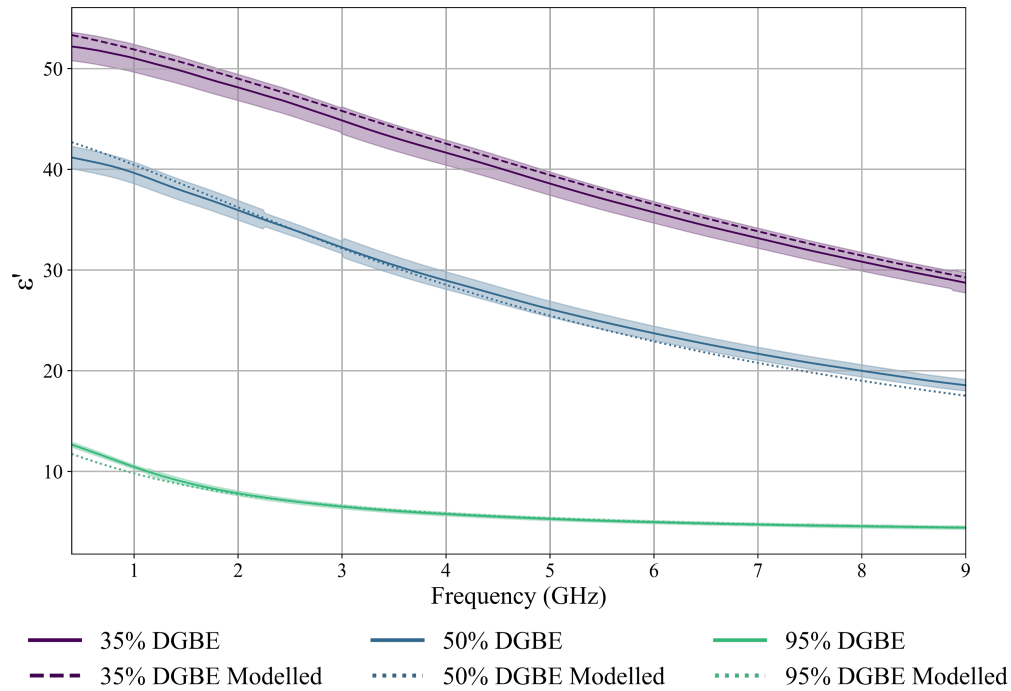


Figure 4.2: The modelled (dashed line) and measured (solid line) permittivity of 35%, 50%, and 95% DGBE solutions, illustrating the ability of the mathematical models to predict the permittivity of the solutions.

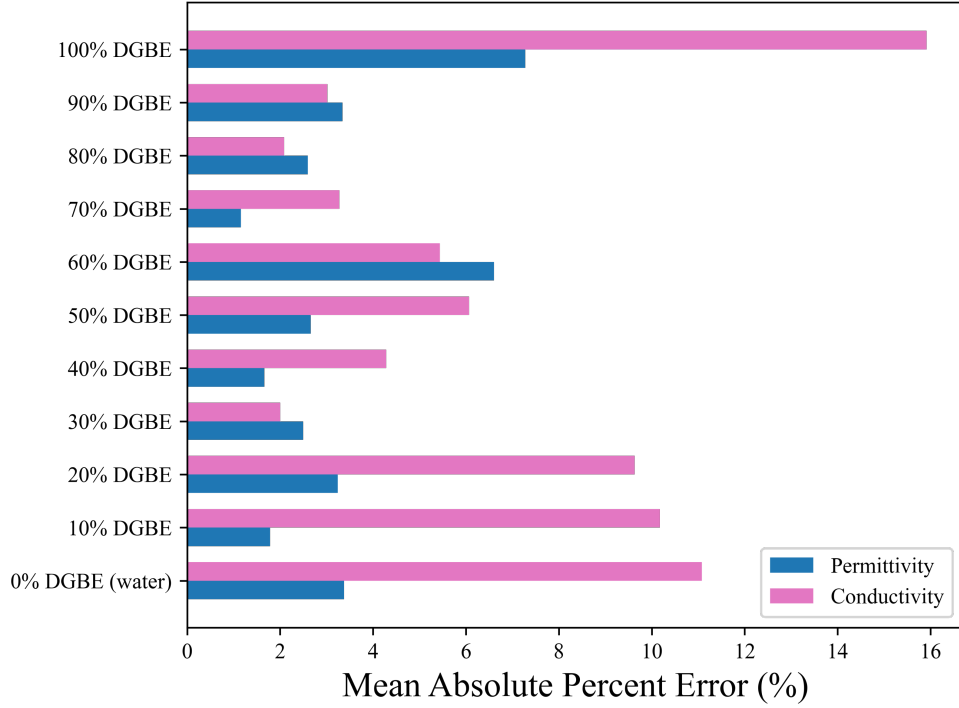


Figure 4.3: The MAPE between the modelled Cole-Cole properties using Eqs. 4.3-4.6 and 4.1 from 0-100%, and the measured data. The average MAPE of the permittivity and conductivity over all DGBE concentrations are 3.3% and 6.6%, respectively.

The findings indicate that solutions containing DGBE can simulate a broad spectrum of dielectric properties. These correspond to the documented properties of *ex vivo* tissues in [130]. Consequently, by varying the DGBE concentration in the solution, it is possible to accurately represent the different types of breast tissue experimentally and allow for contrast resolution experiments to be completed.

4.3.2 Stability of Solutions

When kept in sealed containers, all solutions maintained dielectric stability, with variations within the 3% measurement uncertainty; this suggests that water evaporation is the primary influence on the stability of their properties. The stability test results for the DGBE solutions in sealed containers are depicted in Figure 4.4. Similarly, the 30% Triton X-100 solutions and glycerin out-

comes are illustrated in Figure 4.5. Both figures confirm that the substances remain consistent if stored in airtight containers for several months. In particular, glycerin stored in an open container did not show significant physical alterations over 12 weeks, indicating minimal evaporation. However, as shown in Figure 4.6, glycerin's permittivity exhibited considerable fluctuations during this time due to exposure to environmental factors, although there were no noticeable physical changes.

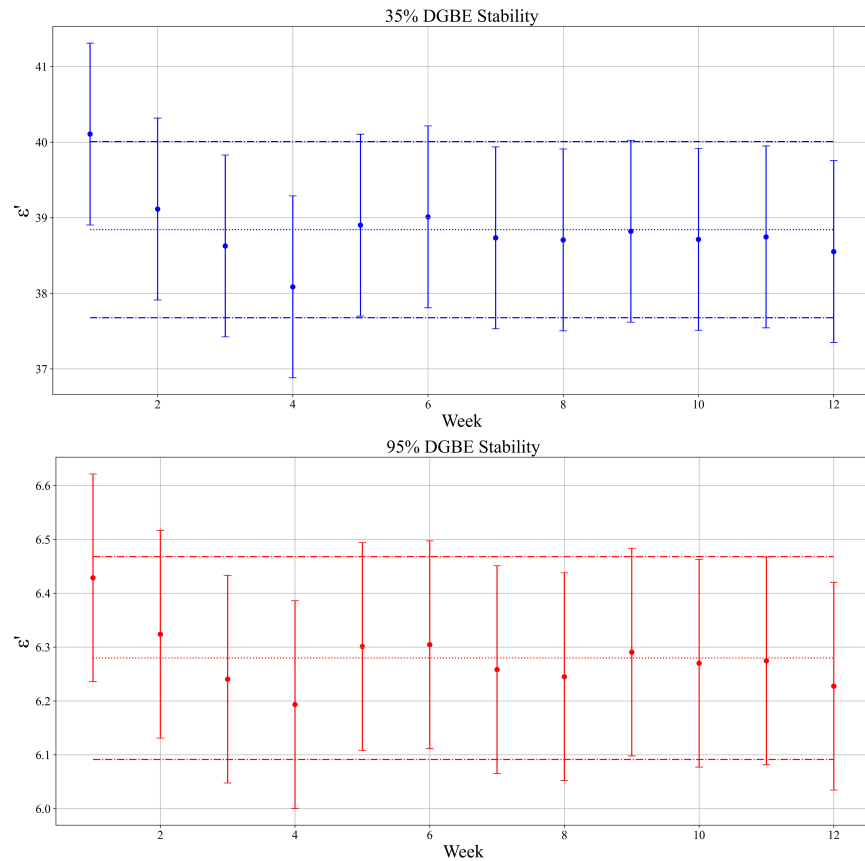


Figure 4.4: The mean permittivity of 35% and 95% DGBE over 12 weeks. The dashed lines show the mean over that time frame and a boundary of $\pm 3\%$ (the measurement uncertainty).

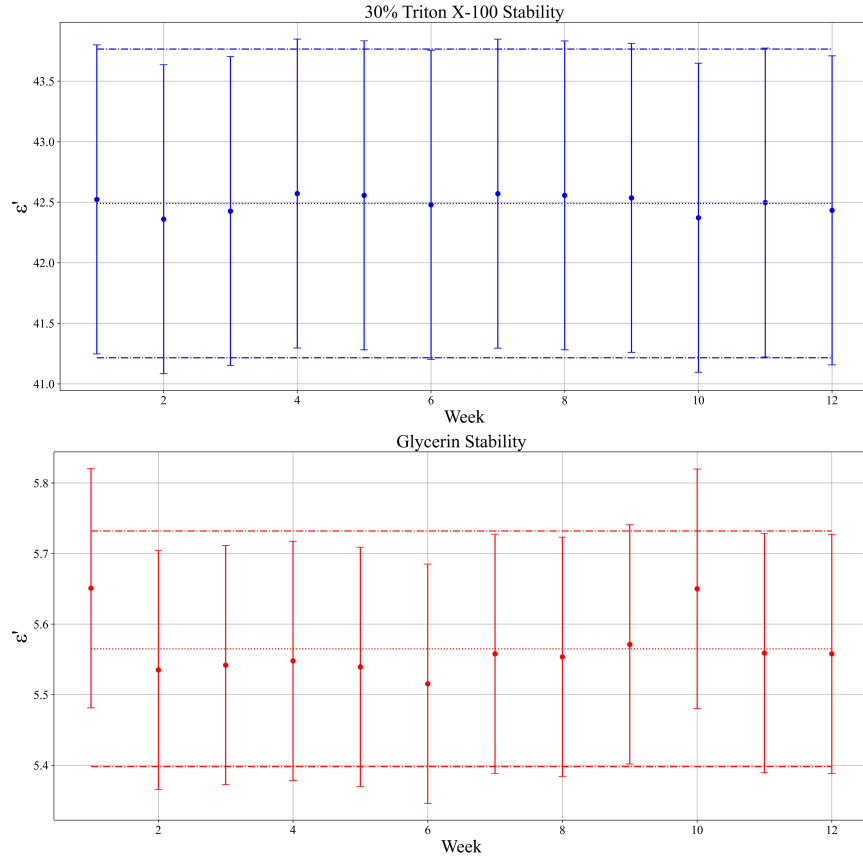


Figure 4.5: The mean permittivity of 30% Triton X-100 solution and glycerin over 12 weeks. The dashed lines show the mean over that time frame and a boundary of $\pm 3\%$ (the measurement uncertainty).

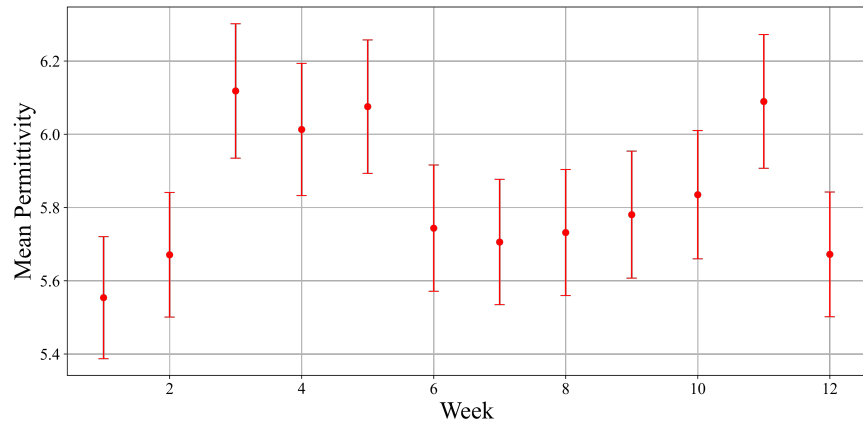


Figure 4.6: The mean permittivity of glycerin stored in an open container over three months. The properties changed significantly during this period.

Investigation of the temperature stability, ranging from approximately 12°C – 30°C , for solu-

tions with concentrations of 35% and 95% DGBE revealed that the dielectric properties of the 95% DGBE solution agreed within $\pm 3\%$ of the properties at room temperature, as shown in Figure 4.7. In contrast, the conductivity of the 35% DGBE solution exhibited notable changes beyond 26°C , as illustrated in Figure 4.8. However, it was observed that the laboratory temperature has not deviated by more than 4°C ; therefore, based on Figure 4.8 and Figure 4.8, it can be said that the properties are stable within (within $\pm 3\%$) of the properties at room temperature (20.9°C) and are therefore known to be within uncertainty during experimentation.

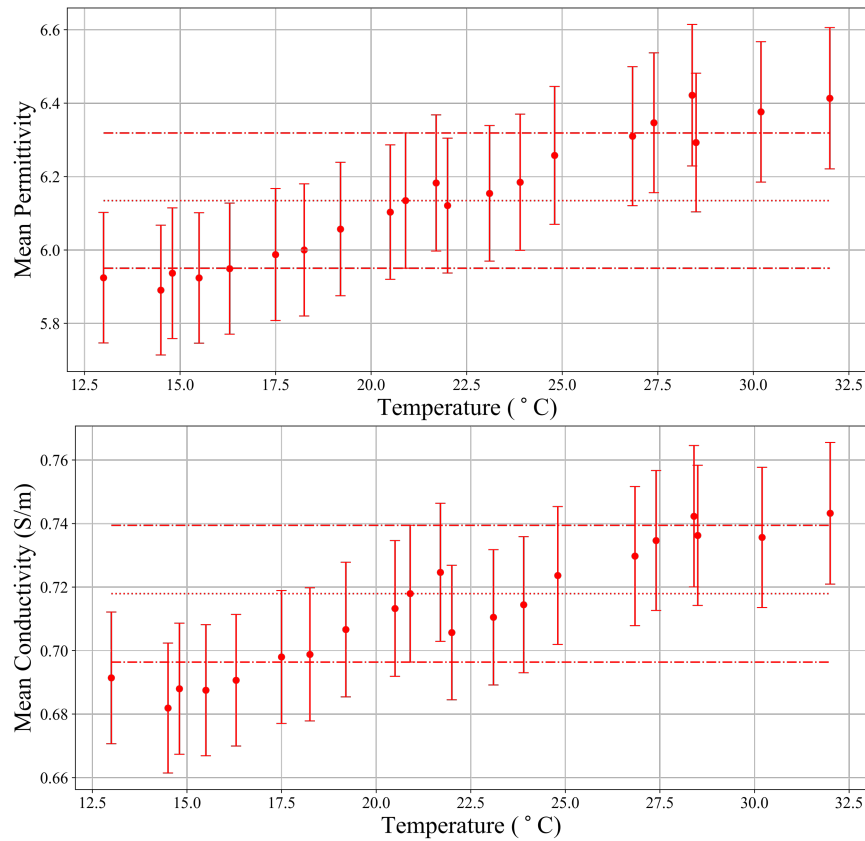


Figure 4.7: The mean permittivity and conductivity of 95% DGBE solutions as a function of temperature. The dashed lines show $\pm 3\%$ of the dielectric properties at room temperature (20.9°C) and were used as a reference to determine agreement.

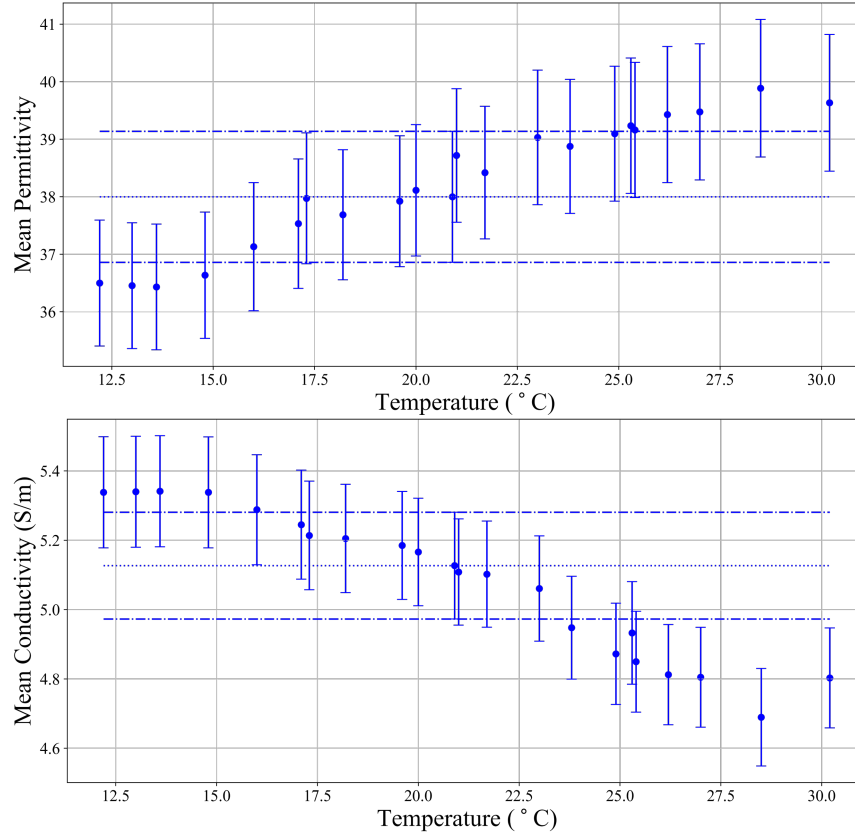


Figure 4.8: The mean permittivity and conductivity of 35% DGBE solutions as a function of temperature. The dashed lines show $\pm 3\%$ of the dielectric properties at room temperature (20.9°C) and were used as a reference to determine agreement. The conductivity will no longer agree with room temperature (20.9°C) at temperatures above 26°C .

4.3.3 Short-term Stability and Restoration of DGBE Solutions

The dielectric properties of the 35% and 95% DGBE solutions stored in open containers were evaluated over three days to determine their short-term stability. The 95% DGBE solution exhibited no notable changes in its properties, as illustrated in Figure 4.9. In contrast, the 35% DGBE solution had significantly reduced permittivity and conductivity, aligning with the hypothesis that water evaporation is responsible for altering properties. Water was reintroduced into the 35% solution to return its properties to the initial state observed before the three-day exposure. The restoration process, as depicted in Figure 4.10, confirms the effectiveness of this method in reinstating the solution's dielectric properties. After three days, the 35% DGBE solution experienced

an 8.85g decrease in mass, attributed to water loss, which was then compensated for in restoration.

Given that DGBE solutions can lose dielectric properties over prolonged exposure due to water evaporation, it is essential to note that when stored in sealed containers, as shown in Figure 4.4, the solutions remain stable for several months. Moreover, the properties can be recovered post-evaporation, as illustrated in Figure 4.10. Therefore, it is practical and recommended that the solutions be reused, provided they are only briefly exposed to air. The solutions may also be checked, and if there is a drop in properties, small amounts of water may be added to restore the properties.

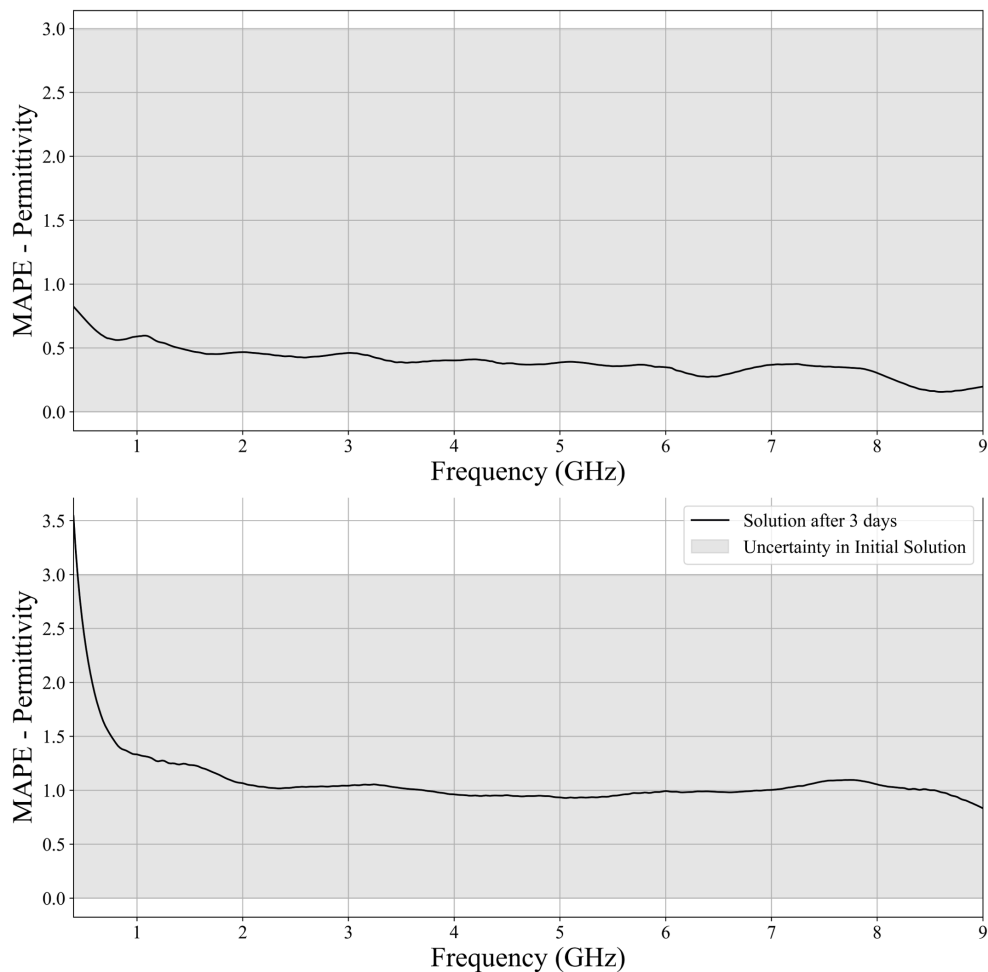


Figure 4.9: The MAPE of the permittivity of a 95% DGBE solution before and after being left in an open container for three days. The solution can be seen to agree with the initial solution, except for the conductivity at the first few lower frequencies (<0.5 GHz).

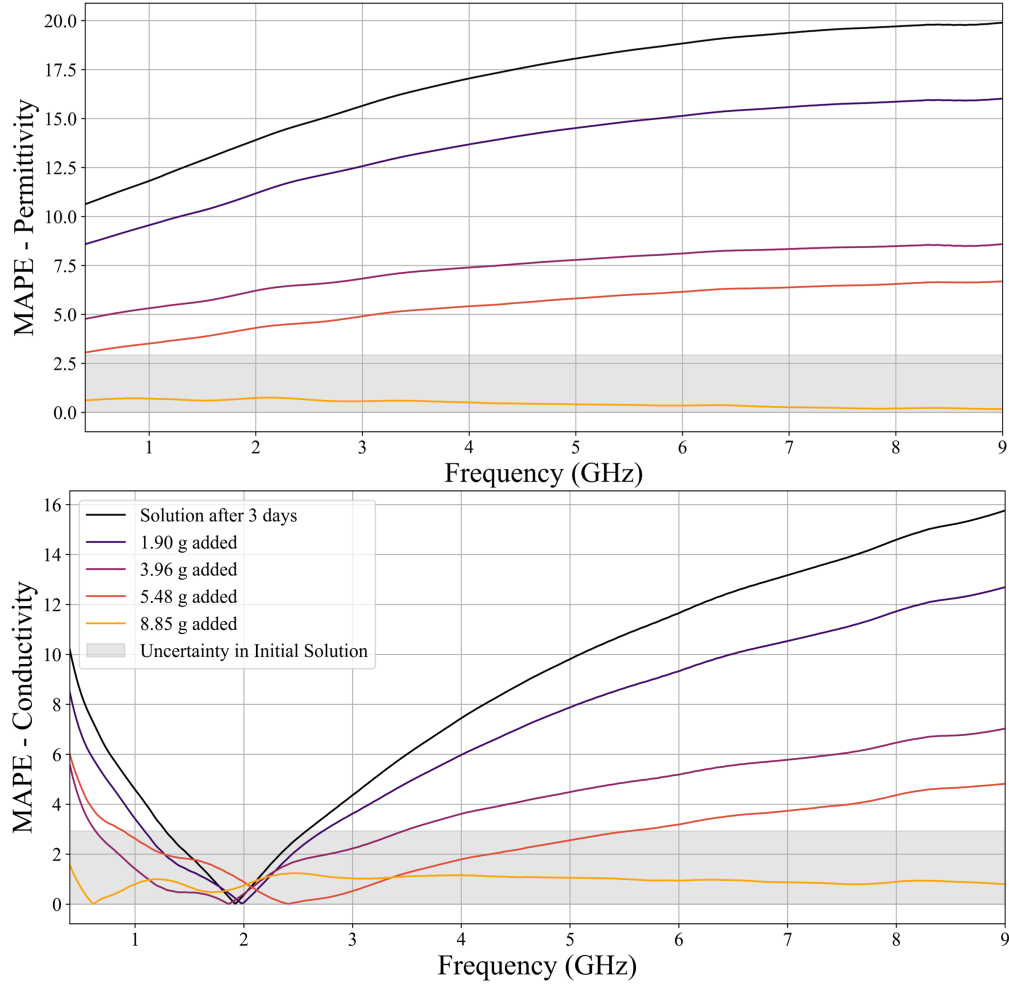


Figure 4.10: The MAPE of the permittivity and conductivity of a 35% DGBE solution after being left in an open container for three days and after adding water. After three days, 8.85 g of water was added to the solution to restore its dielectric properties to those initially observed.

4.3.4 Liquids for Shell-Based Breast Phantoms

Figure 4.11 displays the dielectric properties of the previous liquids, glycerin and a 30% Triton X-100 solution, and the proposed new liquids, 50% and 90% DGBE solutions. For comparison, the dashed lines show the properties of adipose and fibroglandular tissue from [130], while the triangles show the properties of the carbon-rubber adipose-mimicking material in [106]. The MAPE between the liquid and the tissue it has been made to mimic may be seen in Table 4.2.

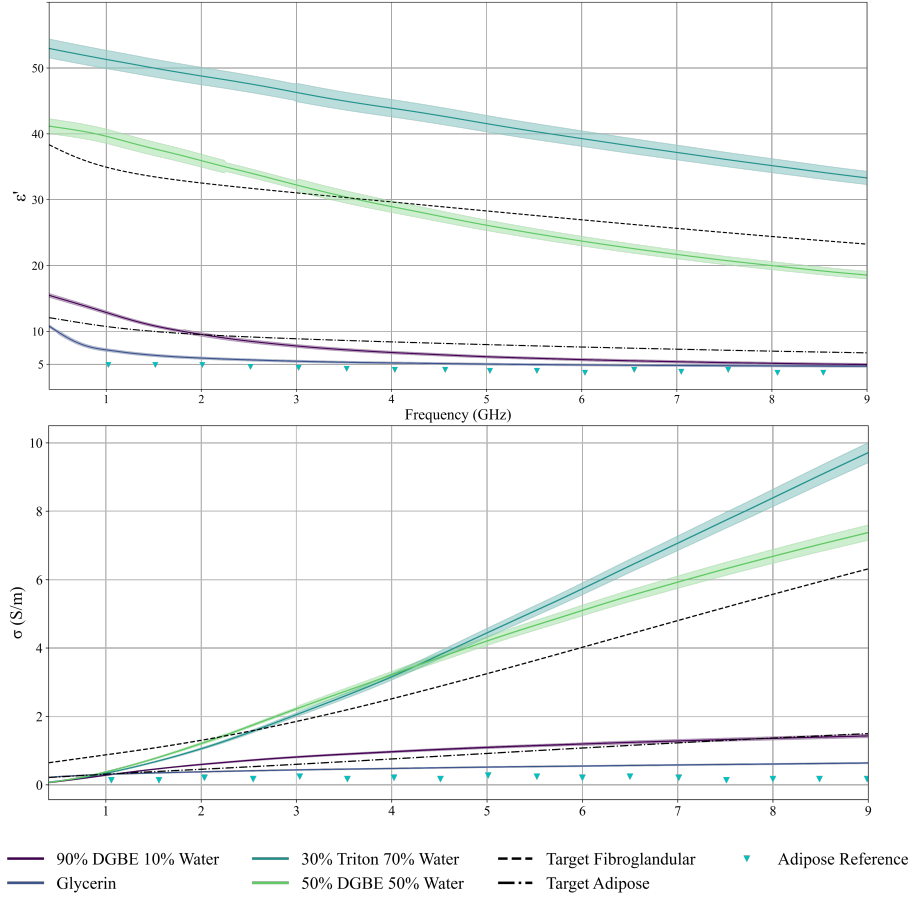


Figure 4.11: The measured dielectric properties of the previously used tissue-mimicking liquids, glycerin and 30% Triton X–100, and the proposed solutions, 50% and 90% DGBE, illustrate the improved properties with the new solutions. The dashed lines are the average tissue properties from *ex vivo* tissue measurements [130], while the adipose reference is the carbon rubber presented in [106].

Table 4.2: Mean absolute percent error between tissue and TMMs

Adipose		
Material	ϵ'	σ
Glycerin	35(2)%	36(2)%
90% DGBE - 10% Water	20(2)%	18(3)%

Fibroglandular		
Material	ϵ'	σ
30% Triton - 70% Water	47(4)%	38(3)%
50% DGBE - 50% Water	11(3)%	27(3)%

The 90% and 50% DGBE solutions enhanced dielectric accuracy compared to the materials previously used, glycerin and 30% Triton X-100. MAPE for adipose-mimicking materials decreased from 35(2)% to 20(2)% for permittivity and 36(2)% to 18(3)% for conductivity. Similarly, the MAPE for fibroglandular-mimicking materials dropped from 47(4)% to 11(3)% for permittivity and from 38(3)% to 27(3)% for conductivity. This reduction in MAPE makes these solutions suitable for representing adipose tissue and fibroglandular tissue in phantom studies over the 0.4–9 GHz frequency range. When comparing the physical properties of the solutions, the DGBE solutions, which have a viscosity similar to water, did not trap air bubbles on the probe during submersion for measurements. In contrast, the glycerin and 30% Triton X-100 solution formed air bubbles on the probe, suggesting that DGBE solutions are less likely to trap air bubbles during experiments. Furthermore, since DGBE solutions are easier to mix than the Triton X-100 solutions, DGBE solutions are a more practical and effective choice for experimental use.

4.4 Conclusion

DGBE solutions were investigated as a potential shell-based TMMs due to their low viscosity and, thus, lower probability of forming air bubbles during experimentation and ease of preparation. The goal was for the liquids to replace the current viscous liquids commonly seen to form air bubbles during experimentation: glycerin and a Triton X-100 solution.

The DGBE solutions covered a wide range of properties over the frequency range of 0.4–9 GHz. The wide range of properties will allow the contrast resolution in different BMS systems to be calculated and compared. The properties of DGBE were modelled using a Cole-Cole model at concentrations of 0–100%. The extracted Cole-Cole parameters were then modelled as a function of concentration, allowing users to obtain theoretical properties at different concentrations. The modelled properties were compared to experimental measurements for concentrations of 35%, 50%, and 95% DGBE and agreement was obtained over the 0.4–9 GHz frequency range, confirming the modelling.

The properties of a 90% DGBE solution improved the MAPE of an adipose-mimicking material compared to glycerin, reducing it (compared to *ex vivo* tissue measurements) from 35(2)% to 20(2)% and 36(2)% to 18(3)% for the permittivity and conductivity, respectively. Similarly, a 50% DGBE solution reduces the MAPE of the Triton X-100 solutions used to mimic fibroglandular tissue from 47(4)% to 11(3)% and 38(3)% to 27(3) % in terms of permittivity and conductivity, respectively. With the wide range of properties shown, concentrations can also be modified to test with a range of properties that can be seen clinically.

Lastly, a 95% DGBE solution was shown to be thermally stable (compared to room temperature) over the temperature range of 12°C–30°C while also showing stability when kept in a closed container for 12 weeks and an open container for three days. Conversely, a 35% DGBE solution was shown to be thermally stable from 12°C–26°C, meaning it is thermally stable at room temperature. The solution was also shown to be dielectrically stable over 12 weeks when stored in a closed container, and when left in an open container for three days, the drop in dielectric properties could be restored by adding water equal to the loss in mass over this time. Overall, this leads to the conclusion that the solutions can be stored at room temperature and used repeatedly if the properties are monitored, and water is added when a drop in properties is observed.

5. Utilization of 3D-Printing for Quality Assurance and Control in Breast Microwave Sensing

This chapter utilizes 3D-printing technologies for use in BMS quality assurance and control. The 3D-printed devices are used in data-driven methods for correcting positioning errors and determining different image quality metrics across different BMS systems.

5.1 Introduction

In BMS, image quality metrics have typically been the image contrast, defined by the signal-to-background ratio, or the accuracy, defined by the localization error [38]. However, traditional image quality metrics, such as signal-to-noise and resolution, have been neglected [38, 132]. These metrics are important when analyzing a system's image quality or identifying how well an image will allow a medical professional to accomplish their goal [133].

It is essential to evaluate the effectiveness of a reconstruction algorithm based on how well it can locate a tumour. Ideally, the measured position of a tumour within the breast is in the expected position. However, in experimental phantom studies, uncertainties in the expected tumour position require a margin of error. Current work in BMS has not adequately defined how large this margin should be; some work even defined an accurate localization when the tumour was reconstructed in the same quadrant of the breast as the expected location [134]. The Localization Error (LE) is a qualitative accuracy metric that can be used and is defined as

$$LE = |\vec{r}_{\text{meas}} - \vec{r}_{\text{img}}|, \quad (5.1)$$

where $\vec{r}_{\text{meas}}$ and $\vec{r}_{\text{img}}$ are the measured and reconstructed tumour locations, respectively. A visualization of the LE can be seen in Figure 5.1

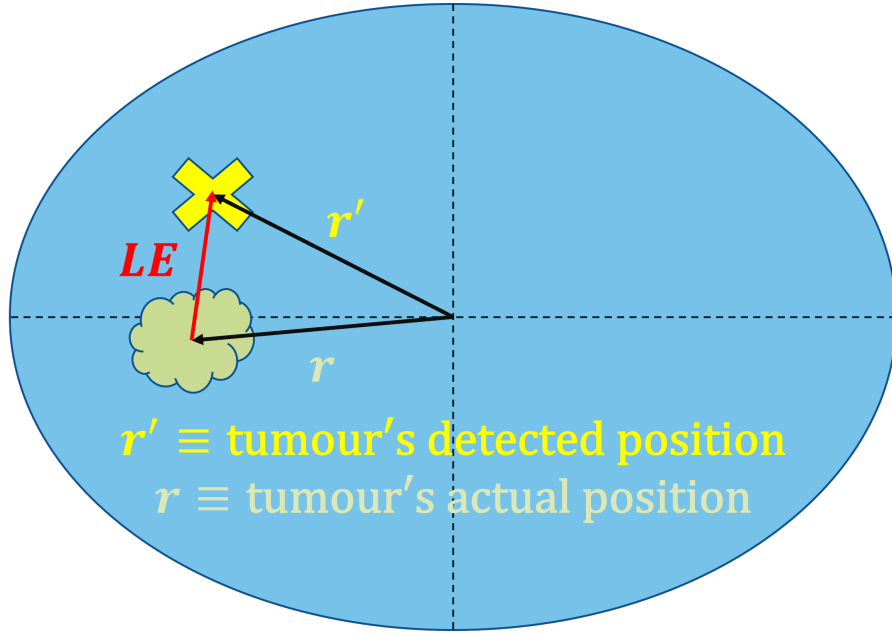


Figure 5.1: The localization error is defined as the positional difference between the image-extracted position and the expected position of the tumour within the breast.

While the localization error is a good measurement of the reconstruction algorithm's accuracy, the literature has not adequately considered other factors that may contribute to the error. Systematic errors can arise when setting up a system. These can arise from rotations and translations between the Object Under Test (OUT) and the antenna reference frame, as shown in Figure 5.2. Furthermore, random positioning errors can occur when placing a tumour in the OUT reference frame. Both of these factors will contribute to the error. Unlike clinical imaging modalities, BMS has yet to describe QA for these setup errors. In x-ray CT, guidelines have been established to minimize these errors by the American Association of Physicists in Medicine (AAPM) Task Group 179 [135].

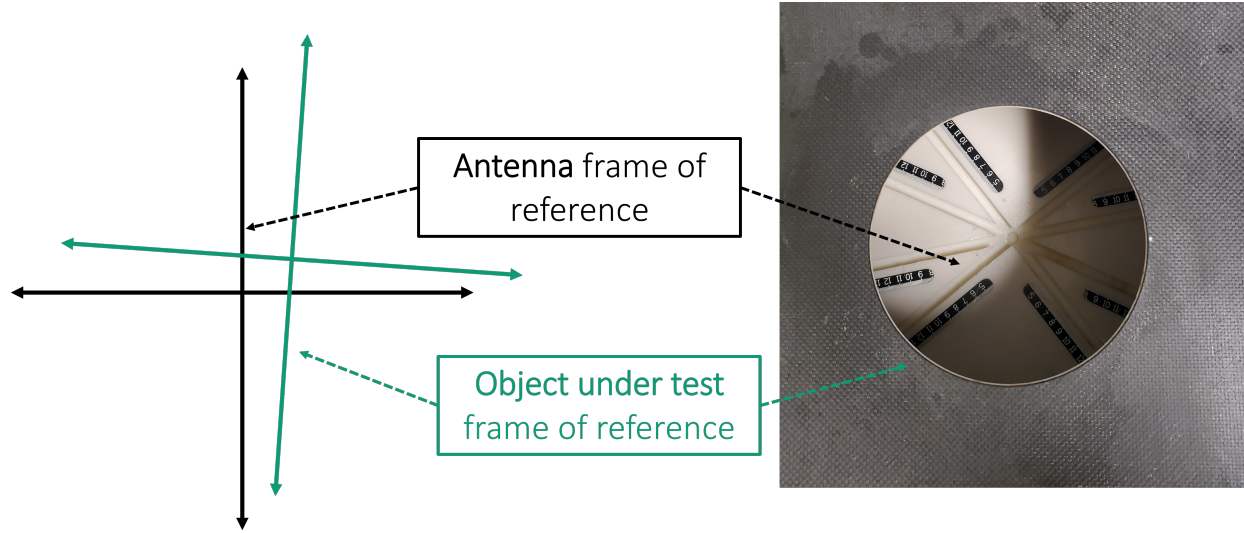


Figure 5.2: A visualization of the difference in the antenna and object under test reference frames that may arise when setting up a breast microwave sensing system.

This chapter will focus on minimizing positioning errors in BMS when positioning a phantom and a tumour. It will present a data-driven method for measuring the position error and, using regression, correct for the errors found. Furthermore, devices for measuring spatial and contrast resolution and noise with a pre-clinical BMS system will be presented.

5.2 Methods

5.2.1 Breast Microwave Sensing Systems

Pre-clinical System

A pre-clinical BMS system was used in all data collection. The system used a bed-based design and is described in [126]. It uses a two-port VNA (C1209, Copper Mountain Technologies, IN, USA) with two horn antennas fixed to a rotating platform. All experiments described were monostatic (S_{11}) measurements over the frequency range of 2–9 GHz. The antenna platform is placed under the breast aperture and must be aligned manually, leading to potential systematic

errors. The system can be seen in Figure 5.3.



Figure 5.3: Bed-styled BMS system.

Intermediate Bench-Top System

An intermediate, multi-static bench-top system was used for comparison. This bench-top prototype was created for experimental research and analysis in a lab environment, with the added benefit of being easily transportable thanks to its compact size. It features a fixed array of 24 Vivaldi antennas, eliminating the need for mechanical movement. The system employs a two-port VNA (Planar 804/1, Copper Mountain Technologies, IN, USA) and an electromechanical 2×24 switch matrix to connect to the antenna array. Operating within a frequency range of 700 MHz to 8 GHz, this system can perform S_{nm} measurements, though only S_{11} data are utilized for this comparison. The system can be seen in Figure 5.4.



Figure 5.4: Intermediate bench-top BMS system.

Portable System

A prototype portable system designed to minimize cost, size, and system complexity was also used for comparison, and the design can be found in [136], but with newer nanoVNAs. The system includes 24 patch antennas arranged in a cylindrical chamber with a 20 cm diameter. Twenty-four compact, cost-effective NanoVNAs are attached radially near each antenna. Currently set up to record only the S_{11} scattering parameter, this multi-static array will enable future multi-port S -parameter measurements. The system scans from 600 MHz to 4.4 GHz. The system can be seen in Figure 5.5.

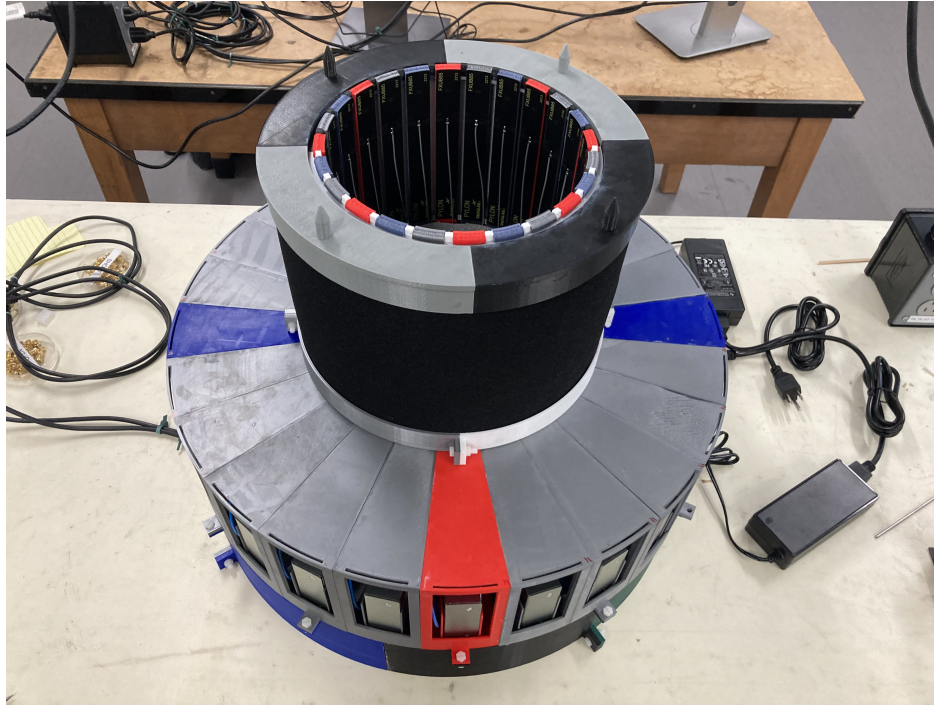


Figure 5.5: Portable BMS system.

Comparison of Systems

A comparison of the three systems is summarized in Table 5.1

Table 5.1: Comparison of the three BMS systems used for comparison.

	Bed-System	Intermediate System	Portable System
Frequency Range	2.0–9.0 GHz	0.7–8.0 GHz	0.6–4.4 GHz
Bandwidth	7.0 GHz	7.3 GHz	3.8 GHz
Number of Frequencies	1001	1001	381
IF Bandwidth	10 kHz	10 kHz	10 kHz
Output Power	15 dBm	10 dBm	-13 dBm
Multi-static Array	✗	✓	✓
Mechanical Motion	✓	✗	✗
Number of Antennas	2	24	24
Number of S-parameter Measurements	72	24	24
Type of Antenna	Horn	Vivaldi	Patch

5.2.2 Positioning Device and Data Collection

A phantom positioning device (Figure 5.6) was designed and fabricated to allow for a repeatable and controllable position of the OUT. The device was initially designed to work with the pre-clinical imaging system presented in [126]. The device was 3D-printed using Polylactic acid (PLA), is capable of centring the phantoms presented in [71], can hold 2-3 mm rods vertically, and can measure positions in up to 0.5 mm resolution, i.e., a radial positioning uncertainty of $\sigma_r = \pm 0.71$ mm. The fabricated device is shown in Figure 5.6, and more information on the design is given in Appendix A.1.

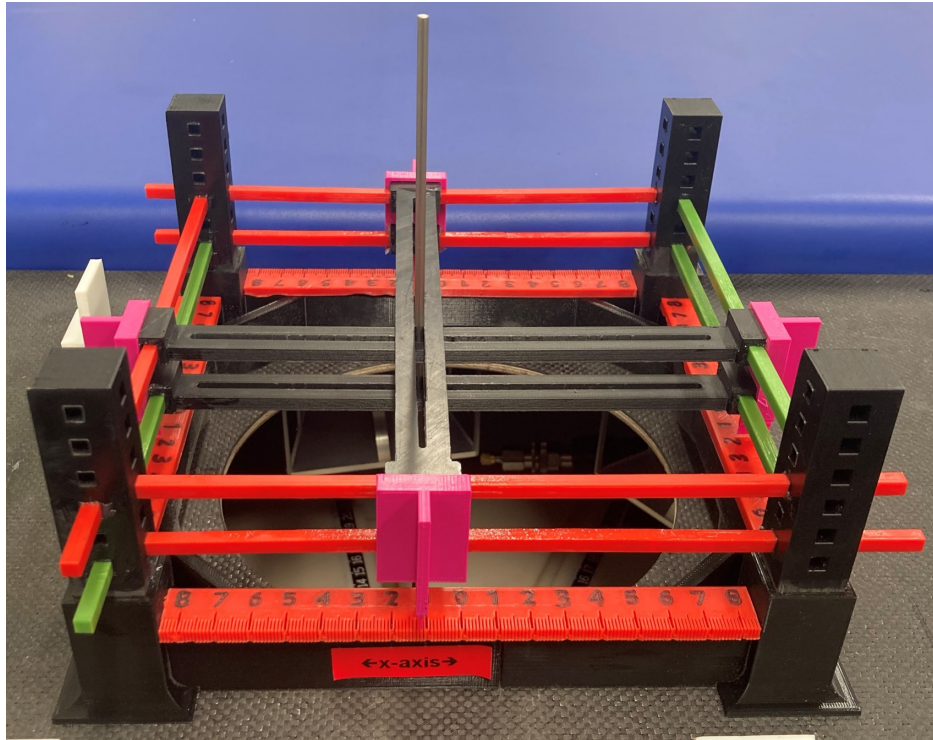


Figure 5.6: The fabricated phantom positioning device.

One hundred and thirteen scans of a 3 mm steel rod were taken, with the rod, at this moment referred to as the object or OUT, assumed to be a point-like source. The object was placed diagonally along the x- and y-axis. The positions were measured in the OUT reference frame. The positions were taken from the rulers and defined as the observed position (x_0, y_0) .

5.2.3 Extraction and Minimization of the Positioning Errors

From the collected data, the positions of the point-like sources were extracted using two methods:

1. Image-space extraction,
2. Time-domain extraction.

The Frequency-Domain (FD) data was reconstructed using the DAS beamformer for the image-space extraction. The beamformer will reconstruct an image in the spatial domain from the S-parameters using

$$I(\mathbf{r}) = \left| \sum_{f=1}^{N_f} \sum_{n=1}^{N_a} \sum_{m=n}^{N_a} S_{n,m}(f) \exp \left(2\pi j f \frac{|r - r_n| + |r - r_m|}{v} \right) \right|^2, \quad (5.2)$$

where $I(r)$ is the image signal intensity at r , $S_{n,m}$ is the S-parameter measured from the signal sent from $\mathbf{r}_m$ and received at $\mathbf{r}_n$, N_f is the number of frequency points, v is the propagation speed, in this case $v = c$, the speed of light, and N_a is the number of ports. In this case, S_{11} was the S-parameter measured and analyzed, i.e., $\mathbf{r}_m = \mathbf{r}_n$. The (x, y) positions were then found by extracting the position of the Center of Mass (CoM) around the maximum signal intensity in the reconstructed image.

For the Time-Domain (TD) method, the FD data was converted to the TD using the Inverse Chirp-Z Transform (ICZT) [137]. The spatial positions were then extracted to fit a model to the measured data. The time delay of the out to the antenna at an angle of ϕ was modelled by

$$t_{\text{pred}} = \frac{2}{c} \sqrt{\rho_a^2 + \rho_t^2 - 2\rho_a\rho_t \cos(\phi_a - \phi_t)}, \quad (5.3)$$

where c is the speed of light in air, ρ_a is the radial position of the antenna, and ρ_t and ϕ_t are the

polar coordinates of the point-like source.

A Levenberg-Marquardt least-squares regression algorithm was used to determine the positions of the point-like source by fitting the modelled time delay, t_{pred} , to the measured, t_{meas} ,

$$\{\rho_t, \phi_t\} = \underset{\{\rho_t, \phi_t\}}{\operatorname{argmin}} \sum (t_{pred} - t_{meas})^2; t_{meas} = \underset{t}{\operatorname{argmax}} |S_{11}(t)|, \quad (5.4)$$

where the maximum response of the TD data gave the measured response time, the radial coordinates of the rods were converted to (x, y) positions.

Rotation and translation can then be used to transform the extracted (x, y) positions to the OUT reference frame,

$$\begin{pmatrix} x_i \\ y_i \end{pmatrix} = \begin{pmatrix} \cos(\delta\phi) & -\sin(\delta\phi) \\ \sin(\delta\phi) & -\cos(\delta\phi) \end{pmatrix} \begin{pmatrix} x \\ y \end{pmatrix} + \begin{pmatrix} \delta x \\ \delta y \end{pmatrix}. \quad (5.5)$$

where (x_i, y_i) are the extracted positions in the OUT reference frame, δx and δy are the translation errors, and $\delta\phi$ is the rotational error. However, the transformation assumes no random positioning errors. The systematic positioning errors can then be found through least-squares regression,

$$\{\delta x, \delta y, \delta\phi\} = \underset{\{\delta x, \delta y, \delta\phi\}}{\operatorname{argmin}} \left(\sum (x_0 - x_i)^2 + \sum (y_0 - y_i)^2 \right). \quad (5.6)$$

Applying the extracted rotational and transition errors to Equation 5.9 will then apply a correction. The random positioning errors, or positioning uncertainty, were then defined as the 95% confidence interval, denoted $2\sigma_r$.

5.2.4 Spatial Resolution Data Collection

The phantom positioning device was used to center a spatial resolution device, which was used to collect data from 3 mm steel rods placed at varying distances from each other. The rods

were supported in holes which started at (0, 0) and spiralled outward with the minimum wall-to-wall separation of each point being 1 mm and the center-to-center spacing starting at 4.4 mm and increasing by increments of 1 mm. The spiral design of the holes allows for more holes to be present and for more spacing permutations. A secondary holder was placed 40 mm below the first to ensure each rod was held vertically. Figure 5.7 shows a photo of the device used in the experiment.

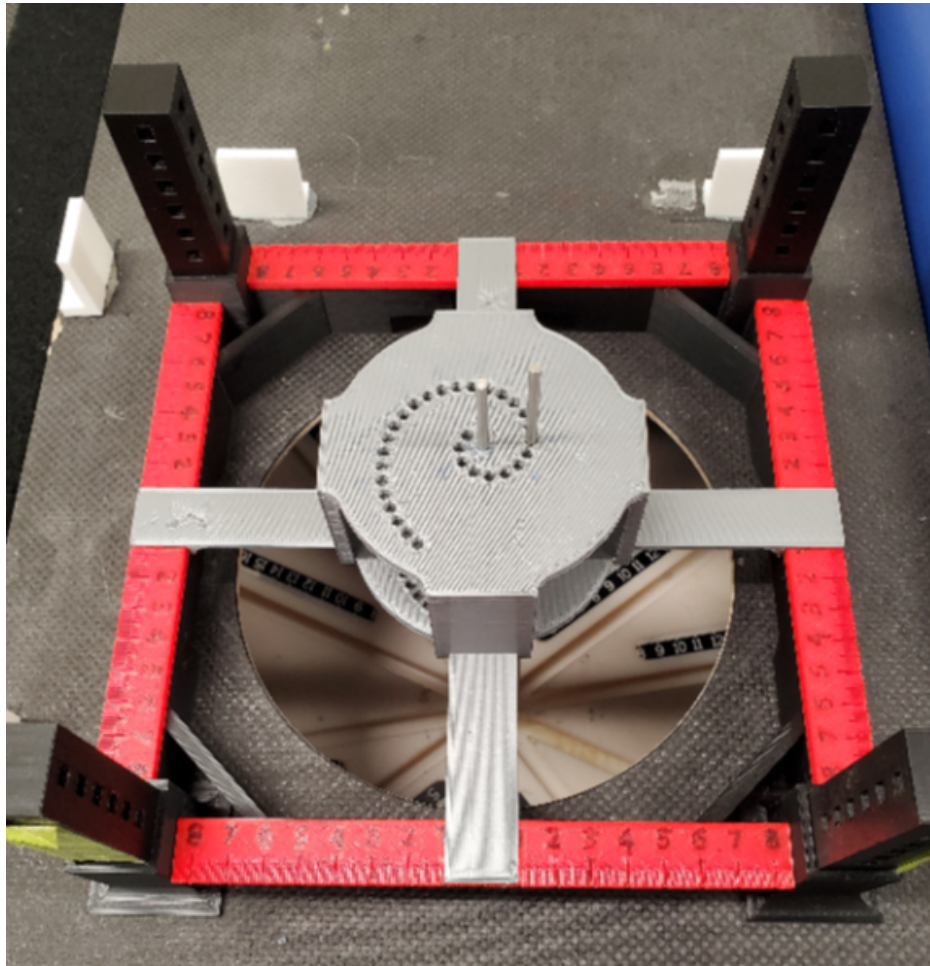


Figure 5.7: The spatial positioning device attached to the phantom positioning device within a pre-clinical BMS system. The device is holding two 3 mm steel rods for spatial resolution measurements.

Measurements were made with the first rod (point-like source) in the center hole while the other rod was moved along the spiral for each scan. The device and data collections allow for a two-

target analysis that determines the best-case resolution with minimal multiple scattering effects, the details of which are outlined in [132, 138]. Furthermore, the positioning device was used in two other BMS systems for comparison: a bench-top styled system and a portable imaging system; descriptions of both can be found in [138]

5.2.5 Contrast Resolution Device

Two OUTs with known dielectric properties are required to make contrast resolution measurements. In chapter 3 and chapter 4, the dielectric properties of various solids and liquids were described. The solid materials were shown to have more variability across samples, making their properties less homogeneous and controlled. Therefore, since DGBE solutions, with properties as shown in Figure 4.1 were used for contrast measurements, containers were required to contain the liquids within the imaging system.

Small cylindrical phantoms constructed with 2 mm PC-CPE walls and 1 mm Stereolithography (SLA) printed with clear resin were compared to determine the optimal container material. Scans of the two materials containing air (nothing), DGBE, and water were taken using the pre-clinical bed system. The median $|S_{11}|$ normalized to the maximum signal intensity were compared as a function of the signal intensity. To select the optimal material, scans were taken of the cylindrical plastic containers containing:

- Air - the only response will be from the plastic.
- A 95% DGBE solution - a low-permittivity liquid.
- An approximately 70% DGBE solution - a higher permittivity solution than the 95% DGBE one, but less than pure water.
- Water - the highest permittivity liquid.

5.2.6 Cylindrical Phantom for Noise Measurements

For any image taken by a system, the image intensities as a function of the spatial position will be given by

$$I(\mathbf{r}) = C(\mathbf{r}) + N, \quad (5.7)$$

where $C(\mathbf{r})$ is the characteristic image intensity, and N is the image noise; i.e., every image will have the actual signal intensities with the intrinsic noise of a system. From Equation 5.7, the image noise can be extracted from the difference between the two images. We can then quantify image noise using the Signal to Noise Ratio (SNR) [138],

$$\text{SNR} = \frac{1}{V} \sum_{\mathbf{r} \in V} \frac{I(\mathbf{r}^{(n)})}{|I(\mathbf{r}^{(n)}) - I(\mathbf{r}^{(n-1)})|}. \quad (5.8)$$

Controlled scans are therefore required to make image noise measurements. A 100 mm in diameter by 120 mm in depth cylindrical phantom was 3D-printed using PC-CPE with a thickness of 2 mm to make such measurements; the fabricated phantom is shown in Figure 5.8. Thirty consecutive phantom scans were taken - the phantom was filled with a 95% DGBE solution. The phantom was positioned in the center of the imaging system using the positioning device and was not further moved during the measurements.



Figure 5.8: The fabricated cylindrical phantom used in the image noise measurements.

5.3 Results and Discussion

5.3.1 Positioning Correction and Error Minimization

Table 5.2 shows the systematic positioning errors obtained using the two methods. The errors were then applied in Equation 5.9 to correct the errors. Figures 5.9 and 5.10 display the observed positions (black), extraction positions (pink) and corrected positions (blue), illustrating that the positional errors can be corrected.

Table 5.2: Systematic positioning errors use delay-and-sum (DAS) and time-domain (TD) positioning extraction methods.

Method	δx	δy	$\delta \phi$
DAS	$(-2.46 \pm 0.07) \text{ mm}$	$(-2.76 \pm 0.07) \text{ mm}$	$(2.8 \pm 0.1)^\circ$
TD	$(-2.68 \pm 0.07) \text{ mm}$	$(-2.84 \pm 0.07) \text{ mm}$	$(3.0 \pm 0.1)^\circ$

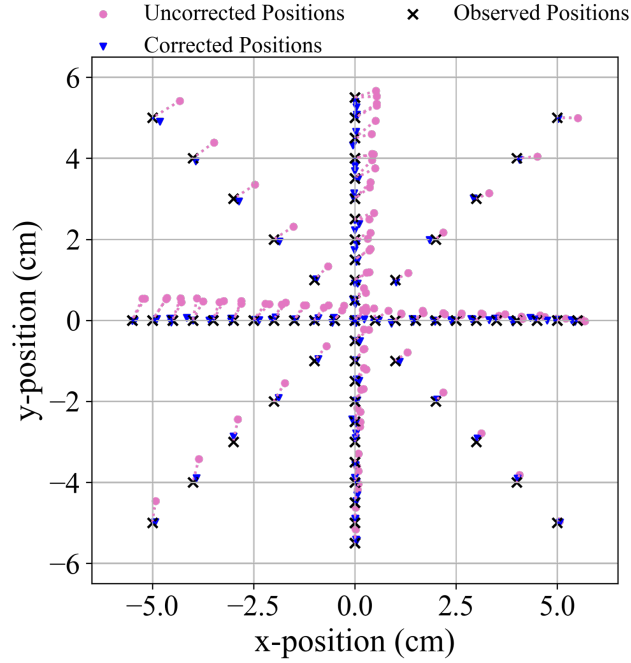


Figure 5.9: DAS method: observed, uncorrected, and corrected rod positions.

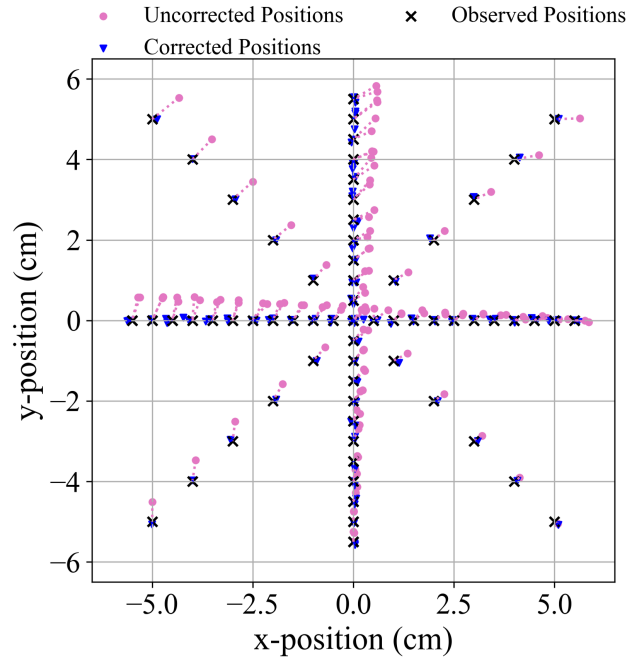


Figure 5.10: TD method: observed, uncorrected, and corrected rod positions.

Both methods obtained similar corrections (Table 5.2), and both reduced the positioning errors (Figure 5.11, Table 5.3). However, even though Table 5.3 shows that both methods had similar

LE and random uncertainty in the position, the TD method used 72 data points from the 72 antenna positions, compared to a single point in the image-based method, making it more accurate. Furthermore, additional errors can be introduced by the DAS reconstruction method itself.

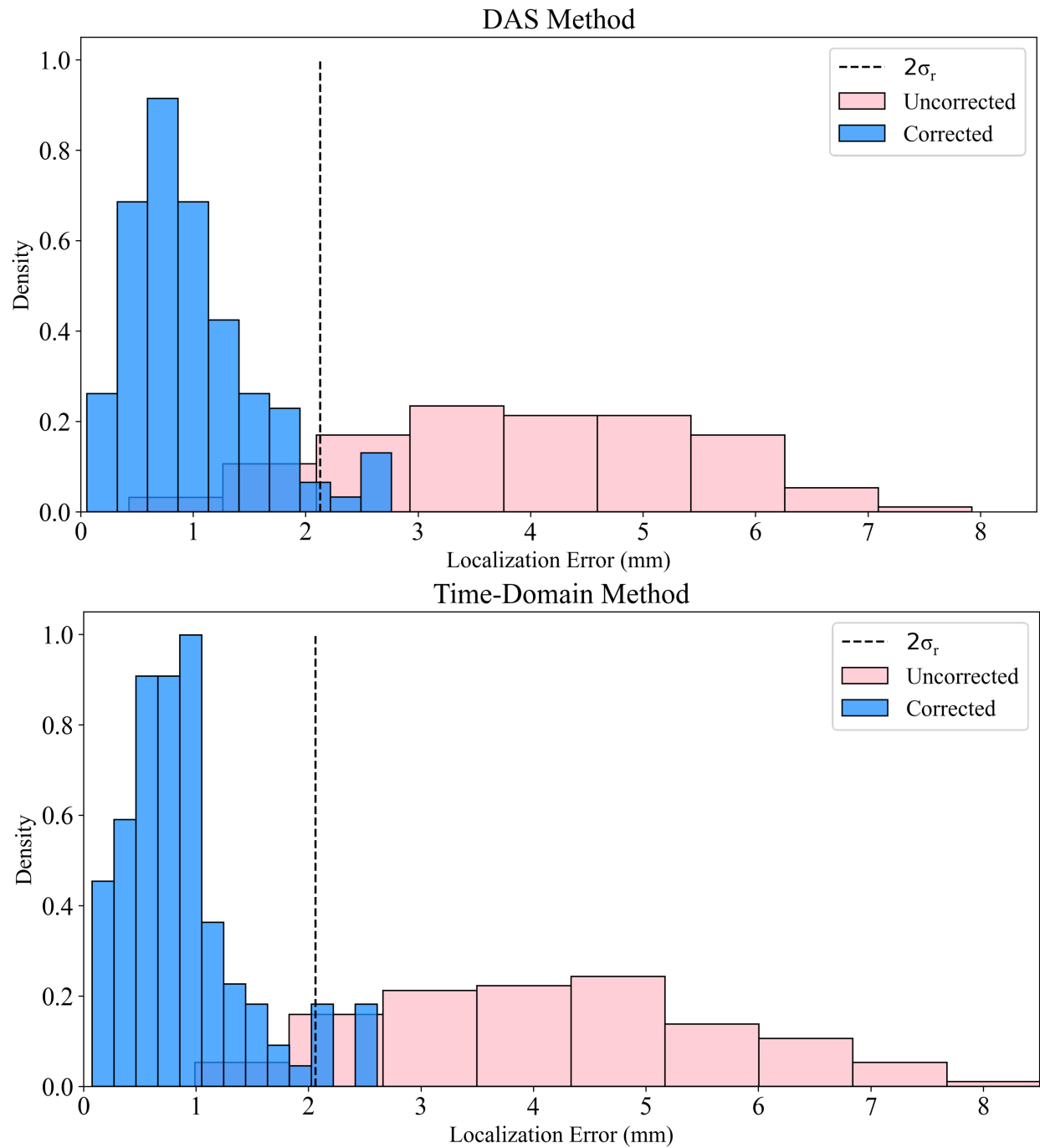


Figure 5.11: Histogram of the localization errors before and after correction using the image-space and TD methods

Table 5.3: Comparison of the impacts of the systematic correction for both the TD and DAS methods.

Method		$2\sigma_r$	mean LE	max LE
DAS	(uncorrected)	± 6.37 mm	3.97 mm	7.9 mm
	(corrected)	± 2.13 mm	0.88 mm	2.7 mm
TD	(uncorrected)	± 7.00 mm	4.20 mm	8.5 mm
	(corrected)	± 2.06 mm	0.88 mm	2.6 mm

Correcting positioning errors in an imaging system reduced the uncertainty from ± 6.37 mm to ± 2.13 mm and ± 7.00 mm to ± 2.06 mm for the DAS and TD methods. The maximum LE was reduced by 5.2 mm for the image-based method and 5.9 mm for the TD method.

A scaling factor was added to Equation 5.9, resulting in

$$\begin{pmatrix} x_i \\ y_i \end{pmatrix} = \begin{pmatrix} \cos(\delta\phi) & -\sin(\delta\phi) \\ \sin(\delta\phi) & -\cos(\delta\phi) \end{pmatrix} \begin{pmatrix} \alpha x \\ \alpha y \end{pmatrix} + \begin{pmatrix} \delta x \\ \delta y \end{pmatrix}, \alpha = a + 1. \quad (5.9)$$

The scaling factor was statistically significant, suggesting stretching in the translations. A delay from the VNA to the antenna was applied to get the spatial coordinates, but an error in this delay would result in the coordinate system requiring a scaling factor. While this identified that a TD correction needed to be applied to more accurately calculate the delay from the VNA to the antenna, a phase delay should preferably be used as this applies a correction as a function of frequency as the phase center of the antenna is frequency-dependent.

Table 5.4: Systematic positioning errors using the delay-and-sum (DAS) and time-domain (TD) positioning extraction methods with the scaling factor included.

Method	δx	δy	$\delta\phi$	a
DAS	(-2.46 ± 0.07) mm	(-2.76 ± 0.07) mm	$(2.8 \pm 0.1)^\circ$	$(-13 \pm 2) \times 10^{-3}$
TD	(-2.68 ± 0.07) mm	(-2.84 ± 0.07) mm	$(3.0 \pm 0.1)^\circ$	$(9 \pm 2) \times 10^{-3}$

One hundred and thirteen measurements are time-consuming, and therefore, a multi-rod system was 3D-printed with PLA, which is centred by the phantom positioning device. The multi-rod system can be seen in Figure 5.12. To minimize the errors caused during fabrication, the device can be rotated four times on the phantom positioning device. Reconstructions of the four scans can be seen within the image space and TD in Figures 5.13 and 5.14. In Figure 5.14, the five rods are difficult to distinguish, and using the image-based method would make it easier to correct positional errors. The corrections for the one hundred and thirteen-rod measurements and the four scans of the fast correction device show how it may be used for quick calibration in Table 5.5, using the image-based method.

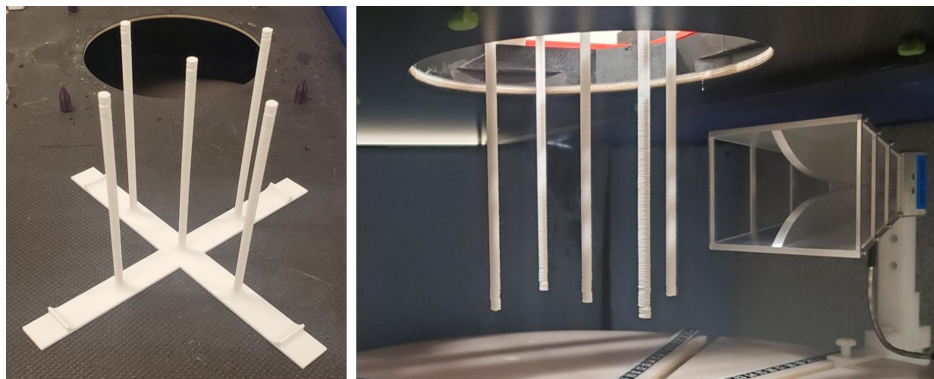


Figure 5.12: The multi-rod positional correction device that can be used for fast calibration of the positioning within a BMS system.

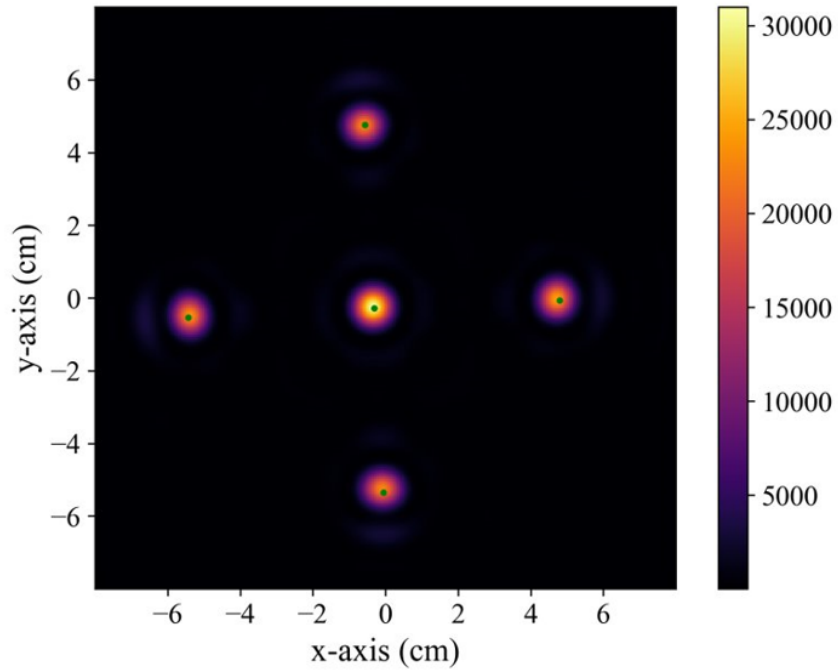


Figure 5.13: Reconstruction of the multi-rod device for positioning error calibration.

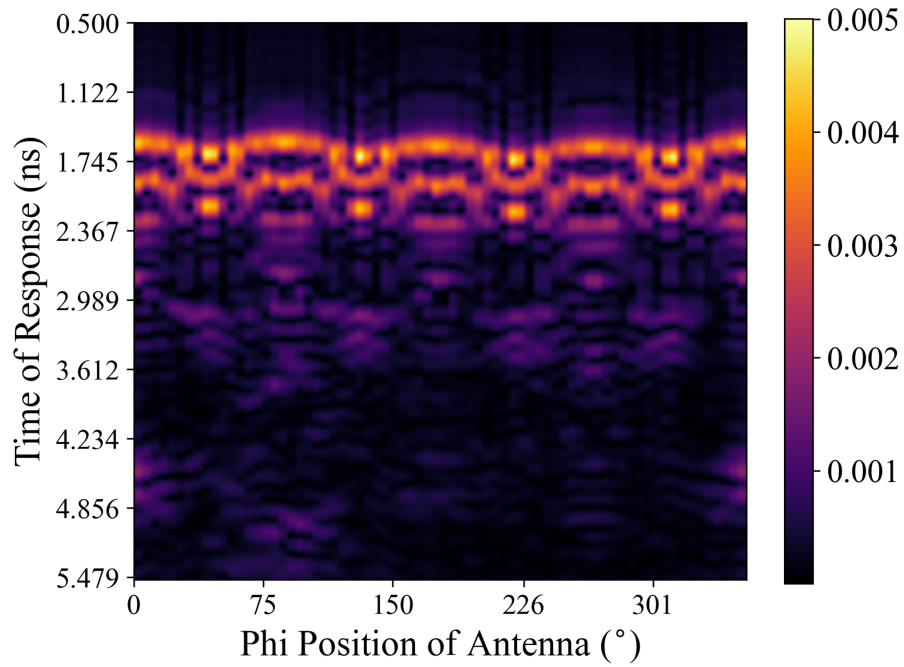


Figure 5.14: Sinogram of the multi-rod device for positioning error. Displaying the difficulty in positional extraction from the device in the TD.

Table 5.5: Systematic positioning errors using the delay-and-sum (DAS) and time-domain (TD) positioning extraction methods with the scaling factor included.

Method	δx	δy	$\delta \phi$	a
113 scans	(-2.46 ± 0.07) mm	(-2.76 ± 0.07) mm	$(2.8 \pm 0.1)^\circ$	$(-13 \pm 2) \times 10^{-3}$
Multi-rod device	(-2.64 ± 0.07) mm	(-2.95 ± 0.07) mm	$(2.8 \pm 0.1)^\circ$	$(18 \pm 2) \times 10^{-3}$

5.3.2 Spatial Resolution

Examples of when the two rods are indistinguishable and distinguishable from [138] can be seen in Figure 5.15. Table 5.6 displays the measured spatial resolutions from the three systems, showing how the higher-spec bed system has the best spatial resolution. The spatial resolution device allows systems to compare spatial resolution with varying hardware or reconstruction algorithms/algorithm parameters, allowing for a proper comparison across BMS systems.

	Spatial Resolution (mm)
Bed System	18 ± 1
Bench-top System	26 ± 1
Portable System	> 34

Table 5.6: The spatial resolution of three systems using the reconstruction of two targets from scans from each system. The results show that the more complex, bed-styled system had the best results [138].

5.3.3 Materials for Contrast Resolution

The median $|S_{11}|$ across different signal intensity thresholds comparing the response of 2 mm PC-CPE and 1 mm SLA holding different fluids can be seen in Figure 5.16. When the cylinder was not filled with material, the SLA had a 60% weaker response than the PC-CPE. Conversely, the SLA always had a stronger response when containing fluids with a higher permittivity. It should expressly be noted that the 70% DGBE solution had a slightly smaller (but distinguishable) response from water. Table 5.7 displays the mean $|S_{11}|$ response for each material normalized to

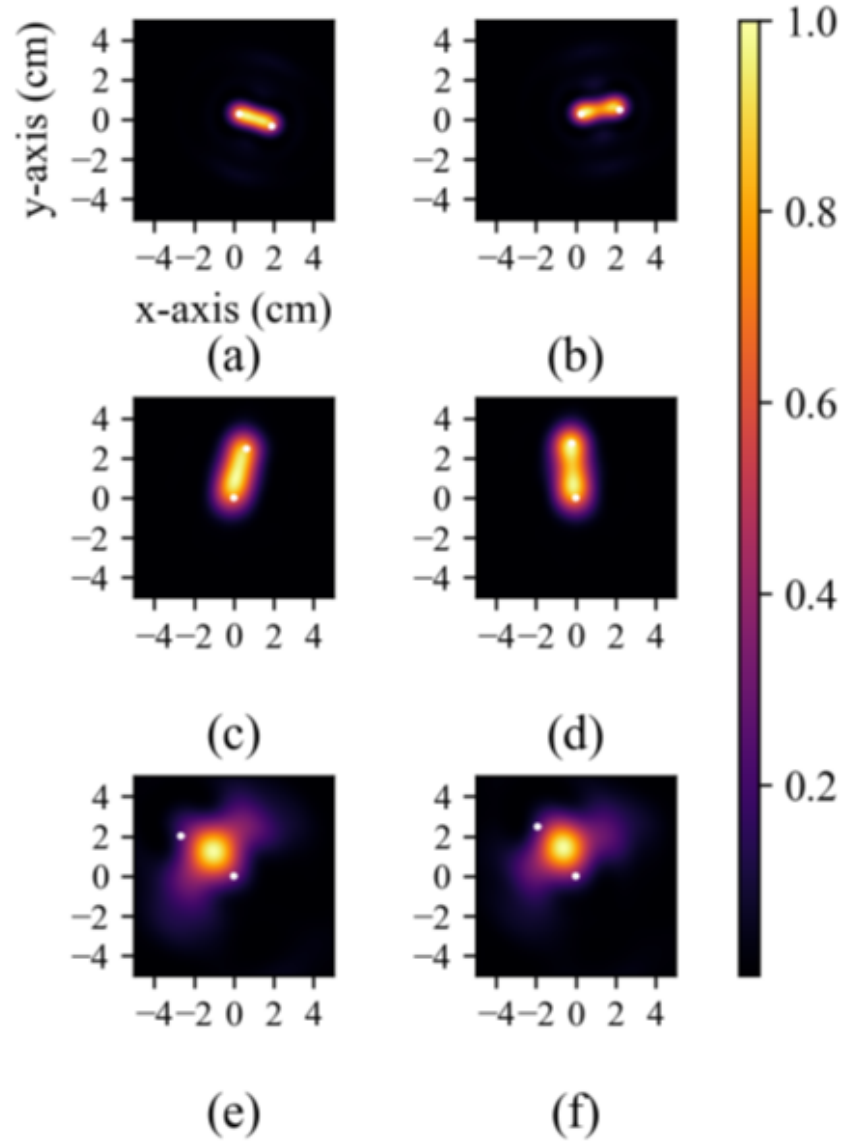


Figure 5.15: Typical reconstructions of the two targets where the targets are indistinguishable in (a), (c), (e), and (f), and distinct, in (b) and (d). The first row was scanned using the bed system, the second with the bench-top system, and the third using a portable system. For the portable system, no two responses were distinct [138].

the response of that material filled with water, i.e., the most significant response of that material.

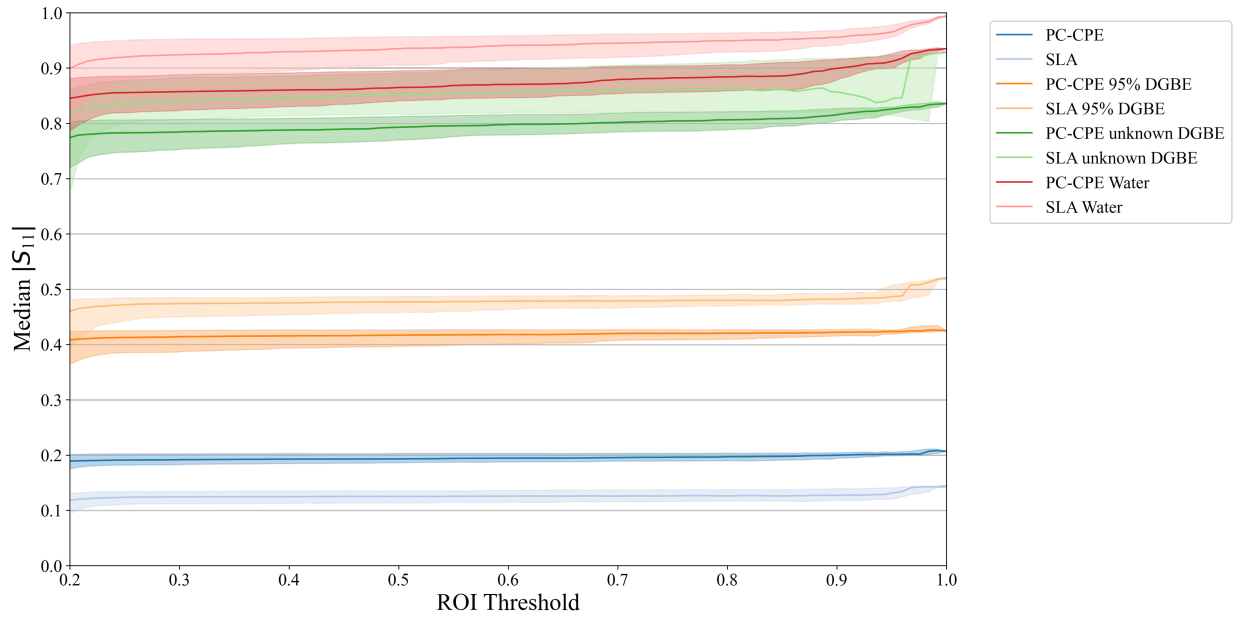


Figure 5.16: The median $|S_{11}|$ across different signal intensity thresholds comparing the response of 2 mm PC-CPE and 1 mm SLA holding different fluids. The results show that SLA containing air has a smaller response (less effect from the plastic material) but a larger response when containing higher permittivity fluids compared to PC-CPE, suggesting that SLA gives a more reliable response of the fluid being scanned.

	Air-filled	95% DGBE	Unknown DGBE
PC-CPE mean $ S_{11} $	0.29	0.52	0.95
SLA mean $ S_{11} $	0.19	0.49	0.90

Table 5.7: The mean $|S_{11}|$ measurement from each material filled with a fluid normalized to the water-filled material.

Figure 5.16 and Table 5.7 illustrate that the SLA has a smaller response when filled with air and a large response when filled with liquid compared to PC-CPE, suggesting SLA gives a more accurate response of the fluid held therein, most likely due to the smaller wall size of the container. The significant response from the PC-CPE allows less of the signal to be reflected from the material itself. In addition, due to the superior 3D-printing resolution of SLA printers compared to FDM printing, the SLA containers could be printed with a 0.5 mm wall thickness. Overall, we recommend that containers for contrast resolution measurements be printed using SLA and that these

containers be further utilized in experimental contrast resolution measurement for BMS systems.

5.3.4 Image Noise Measurement

The results of the image SNR from the three systems are given in Table 5.8. In line with the image resolution, the Bed System has the highest SNR and the lowest variance over 30 scans. Table 5.8 displays the SNR across the three BMS systems, illustrating how the cylindrical phantom can be used to evaluate and compare the noise level of images across BMS system.

	Data SNR
Bed System	26 ± 1
Bench-top System	22 ± 2
Portable System	7.9 ± 0.6

Table 5.8: The SNR was determined from 30 consecutive scans with three different systems. The results show how a cylindrical phantom can compare intrinsic BMS properties across systems.

5.4 Conclusion

A phantom positioning device was designed and fabricated to improve the reproducibility of experimentation in BMS. The device centers phantoms within the aperture of a BMS system to achieve reliable placement. The device had add-ons created and a new methodology was developed to improve quality assurance and control in BMS. The device's design allows it to be used with multiple systems, allowing the comparison of different systems.

A method for improving the QA in BMS was presented. The technique used simple, point-like sources to compare the rods' expected and actual positions in pre-clinical BMS systems. Using regression, the rotational and transitional errors could be corrected; this could be transitioned to any system once set up or used at regular time intervals after set up for QA purposes. The position error was reduced from ± 7.00 mm to ± 2.06 mm and ± 6.37 mm to ± 2.13 mm for the TD and DAS methods. The maximum error was reduced by 5.2 mm for the image-based method and 5.9 mm for

the TD method. This methodology allows the uncertainty in the LE to be determined, providing an estimate of the region around the expected position where the tumour can be classified as correctly located.

A spatial resolution device was designed and fabricated to place point-like sources at known distances from each other within a BMS system. The device was used to compare three different BMS systems: a bed-based system,, a benchtop system,, and a portable system,. The bed system had the best spatial resolution at (18 ± 1) mm, followed by the bench-top system at (26 ± 1) mm, while the portable system could not resolve any of the measurements. The positioning system and spatial resolution device can be jointly used to compare different BMS systems.

PC-CPE and clear SLA resin were compared as containers for DGBE solutions that could be used in contrast resolution measurements. For three different contents - air, 95% DGBE, and a 70% concentration of DGBE, the PC-CPE container had a mean $|S_{11}|$ of 0.29, 0.52, and 0.95, respectively. In contrast, the SLA container had a mean $|S_{11}|$ of 0.19, 0.49, and 0.90, respectively. The lower response of SLA for all measurements, particularly the air-filled scans, suggests that the SLA contributes less to the response from the OUT. These results and the superior print resolution of SLA printers compared to FDM printing leads us to recommend using SLA as the DGBE container for contrast resolution experiments. The positioning device can hold the containers at known positions for accurate experimentation.

Lastly, a cylindrical phantom was designed to be held centred by the positioning device. The cylindrical phantom was filled with 95% DGBE and scanned 30 consecutive times in the three BMS systems that have been previously compared. The bed system was found to have had the best SNR of 26 ± 1 , followed by the bench-top system at 22 ± 2 , with the portable system having the worst results at 7.9 ± 0.6 . The results illustrate the positioning device and phantom's ability to measure noise measurements across three different BMS systems.

Overall, this chapter showed how 3D-printing can fabricate devices for the QA of BMS systems and can be used to measure various image quality metrics. Furthermore, it was shown that the

devices created could be used across multiple BMS systems to compare the metrics when different hardware is used. The devices can also be used to controllably compare the contrast and spatial resolution of varying imaging algorithms.

6. Conclusions

Breast Microwave Sensing (BMS) has been an active area of research over the past forty years as a potential alternative to current methods of breast cancer screening. Both image- and ML-based methods are current research fields for tumour detection. Before testing detection techniques on clinical patient data, the algorithms should be validated with experimental phantom studies. However, the design of these phantoms varies from group to group, with no standard phantoms being available. Furthermore, appropriate quality assurance and control methods are rarely discussed in the literature. This thesis investigates the materials used in shell-based breast phantoms to accurately model the breast. The ideal breast phantom can provide a standard for the field. Additionally, 3D printing can be used to create quality assurance and control devices in BMS. This work addresses the need for standardized testing of algorithms and hardware to be compared adequately across the field.

In Chapter 2, the design of different phantoms used in the field was reviewed, and the pros and cons of each design type, grouped into gel-, solid-, and shell-based designs, were discussed. The ideal phantom was defined to have the following properties:

- **Dielectrically accurate:** the properties of each tissue represent the average and range of dielectric properties seen clinically.
- **Morphologically correct:** the tissues should represent the morphology of the respective tissue, preferably MRI/CT derived.
- **Stable over time:** the properties should remain stable over a reasonable amount of time, i.e., the phantom should not need to be frequently re-fabricated.
- **Reproducible:** the phantom should be able to be recreated by the same research group or

even better, by other research groups for comparison (open access files if CAD design)

- **Complete knowledge of the tissue locations:** knowledge of where the tissue components, i.e., tumour and fibroglandular tissues, are located within the breast; minimal uncertainty in tissue component positions
- **Versatile and diverse:** the more scans from "one" phantom design, the better mix and match adipose + fibroglandular, freedom of tumour positioning, and freedom of fibroglandular positioning. The phantoms should also be able to represent a range of shapes, sizes, and densities.
- **Low cost:** Ideally, it uses lost-cost materials, especially if dielectric properties are unstable over time.

From the review, it was recommended that the carbon-doped materials used in the solid-based phantoms be investigated for use in shell-based materials to improve the shell's dielectrically properties. The combination of the two designs was identified to help the field move closer to achieving an ideal phantom.

Chapter 3 investigated solid-based carbon-doped materials that could replace the current low-permittivity shell material used in shell-based phantoms. Carbon-doped FDM filaments were investigated for anisotropy that would be visible in S_{11} and S_{21} measurements. It was shown that printing CF-PLA in one direction affected the S_{11} by a percent change of 80.3% when rotating from 0-90 degrees. Similarly, CPLA printed vertically, the same way the phantom walls would be printed, had significant change when rotating from 0–90 degrees, with a maximum percent change of 8(1)% and 47(3)% for the S_{11} and S_{21} AUC, respectively. The anisotropy implies that the response of the shell material would vary with the geometry of the phantom, making carbon-doped FDM filaments unsuitable for BMS phantoms.

Carbon-doped resin was also investigated as a potential shell material. The resin was used over the polyurethane used in solid-based phantoms since it has a lower viscosity, making air bubbles

less likely to form in the cured material and making the carbon easier to mix. The resin also cures into a Shore D hardness versus a Shore C hardness, meaning it would have a more controlled shape and would not stretch or tear when containing liquid. 15% graphite - 85% resin was determined to be a suitable adipose-mimicking material that could be used to construct the fibroglandular shells. The MAPE between the shell material and *ex vivo* adipose tissue measurements was reduced from 69(2)% and 99.40(5)% for PC-CPE to 20(4)% and 88(2)% for the permittivity and conductivity, respectively.

Similarly, 35% graphite - 3% CB - 62% resin was determined to be a suitable skin-mimicking shell. The material reduced the MAPE between shell material and *ex vivo* tissue measurement properties from 93.0(4)% and 99.84(2)% for PC-CPE to 30(10)% and 30(30)% in permittivity and conductivity, respectively. Both proposed shell materials had properties comparable to the carbon-polyurethane used in the field. Still, using resin allows for easier mixing, fewer air bubbles, and more suitable mechanical properties for holding liquids.

In Chapter 4, DGBE solutions were investigated to replace glycerin and a 30% Triton X-100 solution currently used to mimic adipose and fibroglandular tissue, respectively. The DGBE solutions were shown to have a wide range of dielectric properties over 0.4–9 GHz. Cole-Cole parameters were extracted from 0–100% DGBE solutions, and mathematical models of the parameters were created as a function of concentration:

$$\epsilon_h = \begin{cases} 2.66x + 3.28, & x \leq 0.53 \\ -3.54x + 6.58, & x > 0.53 \end{cases} \quad (6.1)$$

$$\epsilon_s = -68.13x + 77.96 \quad (6.2)$$

$$\tau = 110.4x^2 - 16.5x + 9.8 \quad (6.3)$$

$$\alpha = \frac{1}{100}(62.4x^3 - 104.6x^2 + 64.3x - 0.3) \quad (6.4)$$

The mathematical model was used to model a 35%, 50%, and 95% DGBE solution and was com-

pared to experimentally measured properties and agreed within the uncertainty of the measurements, validating the model.

For an adipose-mimicking liquid, a 90% DGBE solution was shown to improve the MAPE between the liquid and *ex vivo* tissue measurements from 35(2)% and 36(2)% for glycerin to 20(2)% and 18(3)% in terms of permittivity and conductivity respectively. Similarly, a 50% DGBE solution reduced the MAPE for a fibroglandular-mimicking liquid from 47(4)% and 38(3)% (30% Triton X-100) to 11(3)% and 27(3)% in terms of permittivity and conductivity, respectively. When using a 95% and a 35% DGBE solution, it was shown that the solutions were thermally stable when used around room temperature, were temporally stable when kept in a closed container, and when the dielectric properties dropped due to water evaporation, could be restored to their original properties with the addition of water, meaning the solutions can be reused, reducing waste and cost. Additionally, since the properties of breast tissue vary, the DGBE solution can be varied to study the detection ability of algorithms for different tissue contrast. Furthermore, the DGBE solutions can be used for contrast resolution studies in BMS.

In Chapter 5, 3D-printed devices were used for quality assurance and control, focusing on collecting data for metrics that could be measured and compared between BMS systems. Firstly, a positioning device was created that could consistently place the shell-based phantoms within an imaging system. The positioning device was used to collect a data set of point-like measurements. The data set was used to determine the position of the point-like sources concerning the antennas; these positions were compared to the measured positions on the positioning device, and regression was used to correct systematic errors and reduce the uncertainty in the position. The corrections reduce the random uncertainty in the position in DAS reconstructed images from $2\sigma_r = \pm 6.37$ mm to $2\sigma_r = \pm 2.13$ mm. Similarly, the max LE was reduced in the DAS-reconstructed images from 7.9 mm to 2.7 mm. The uncertainty in the positions gives a region around the reconstructed tumour position that would be acceptable to classify as correctly located.

Various devices were presented in Chapter 5 that can be used to compare different systems. A

spatial resolution device and a cylindrical phantom were fabricated for noise measurements, and data was collected for three different BMS systems. The spatial resolution and SNR were then compared between the three systems, showing how 3D-printed devices could be used to compare hardware differences. SLA was identified as a more suitable container for DGBE solutions for contrast resolution measurements than PC-CPE. The containers could be placed in known positions using the positioning device to compare BMS systems or track temporal changes. Overall, it was shown that 3D printing could be used to create devices that could collect data that can be used for quality assurance and control, which can be compared across various BMS systems.

This thesis proposes several methods by which BMS testing can be improved. The combination of materials in solid- and shell-based phantoms improved the properties of shell materials. The construction of phantoms using these materials can lead to the ideal breast phantom. Researchers could use various breast phantoms to evaluate different BMS systems and algorithms. DGBE was identified as a suitable tissue-mimicking liquid that could be used in various concentrations to test detection capabilities in different contrast scenarios that may be seen clinically. Furthermore, these solutions could be used in non-phantom contrast resolution measurements and to compare BMS systems.

This thesis also shows how 3D-printable devices could provide controllable positions of phantoms and point-like sources and collect data for various image quality metrics, which can also be used to compare system designs and algorithms. Overall, the results show that the shell-based materials can be improved, leading to ideal standardized phantoms in BMS that can give one-to-one comparisons of algorithms. The results also show how 3D-printed devices can be used in quality assurance and control in BMS, an area lacking discussion in the BMS field.

A. Appendix

A.1 Phantom Positioning Device Design

The phantom positioning device consists of four main components:

- **Base**

- The base was designed so that each corner has a hole centred by the pegs in the imaging systems' platform using a peg-and-hole design, allowing the device to be repeatably placed on the system.
- The base has 10 mm removed from a 100 mm radius circle, allowing for the fibroglandular shells in [71] to be centred to the imaging aperture; this allows for the position of the phantom to be known and repeatable.
- Each corner of the base has a square cutout to allow for the post to be placed within it.
- The top of the base has a rail to which the ruler of the device could be clipped.

- **Post and Rails**

- A 100 mm was printed with 5 mm square cutouts placed every 10 mm to hold the rails.
- Rails are held by the post, allowing sliders to move along them and hold the tumour or OUT.
- Multiple rails ensure that the rod/tumour holder will hold any straight objects vertically.

- **Rod/Tumour holder**

- The rod/tumour holder was designed to hold the tumour used in [71] or 2-3 mm metal rods. Two holders are on the device - one for the x- and one for the y-axis.

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- The holder can slide along the rails to allow continuous positioning throughout the device.
 - The holder has a tab that is used to slide it along the rails. The tab drops vertically with a centred 0.5 mm position indicator aligning with the ruler.
 - The x- and y-axis holders are shifted vertically to not interfere with one another while being positioned.
 - A hash-tag-styled clip is slid over the holders to help ensure that the holder is aligned on the top and bottom rail.
 - The holders can be moved to position the OUT up to ± 65 mm along the x- or y-axis.

- **Ruler**

- Designed to have ticks every 0.5 mm and to clip to the base.
- every 5 mm, the tick is longer to allow users to identify the 5 and 10 mm markers.

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