Breast cancer is the second most common type of cancer in the world, and by far, the most frequent cancer among women with an estimate 1.67 million new cancer cases diagnosed in 2012 (25% of all diagnosed cancers in females)1. As with many other types of cancer, early detection and effective treatment are major factors contributing to the long term survival of the patient 2.Mammography is the current standard for the detection of breast cancer. Although this imaging modality has led to a decrease in mortality rates, false positive rates between 60%-70% and a false negative rate of 10%-15% allow room for improvement3. Modalities such as Ultrasound (US) and Magnetic Resonance (MR) imaging can improve the accuracy of diagnosis but MRI is not cost effective as a screening tool and US images require a skilled practitioner to interpret and have not been shown to reduce mortality3.

In the last 20 years, a number of complimentary approaches for breast cancer detection have been proposed. Among these novel techniques, Active Microwave Imaging(AMI) has emerged as a promising imaging technology for breast imaging. AMI is based on the considerable dielectric contrast between healthy and malignant breast tissues, of at least 2 to 1 according the latest in vivo studies 4,5,6,7. This imaging approach may prove to be less harmful and stressful for the patient, since the signals are not ionizing, have a power of less than 10 dBm (which is significantly less than the peak power output of a typical cell phone, which is 27dBm) and do not require breast compression 8, 9. AMI modalities is categorized as Microwave Tomography (MT) or Breast Microwave Radar (BMR). MT forms a dielectric profile using electromagnetic waves of selected microwave frequencies, by solving a nonlinear and ill-posed inverse scattering problem 4, 9. Breast Microwave Radar (BMR) uses Ultra Wide Band (UWB) signals to form a reflectivity map of the scanned region 10, 11. While BMR approaches cannot generate a dielectric map they determine the location of strong scattering signatures which are associated with malignant lesions and are capable of forming high contrast 3D images where mm size inclusions can be resolved 10,11,12,13.MT and BMR have been successfully tested in small-scale clinical trials with patients over the age of 50 4, 5.

During the last six years, a considerable amount of work in the BMR area has been focused on increasing the image quality by optimizing the data acquisition protocols and developing novel image reconstruction methods. Currently, BMR imaging protocols can be divided into monostatic and multistatic scan protocols. In monostatic scan protocols, the same antenna is used to transmit the UWB waveform and record the responses from the breast structures. Multistatic protocols in the other hand, also use different pairs of transmitter and receiver scan locations to illuminate the breast from additional orientations, increasing the information present in the BMR dataset4. Images formed using monostatic datasets usually exhibit a higher focal quality, whereas, multistatic approaches yield datasets with higher sensitivity and specificity. Nevertheless, the majority of BMR image reconstruction is performed using time-domain processing methods that generate images with artifacts and require a considerable amount of priori information, such as the propagation losses in the breast tissue and the length of the signal integration window.

An alternative approach to generate BMR images is by using holography techniques 14. This approach process the spectrum of the collected reflections and transfers it from the spatial-temporal domain where it is originally acquired to the spatial domain where it will be displayed, similarly to k-space reconstruction techniques used in MRI. Previous studies performed by the authors show that holography techniques generate images with similar spatial accuracy, higher SCR, higher contrast and less artifacts when compared to time-shift approaches. The holographic approach described by the authors in the study14 do not require considerably less a priori information in comparison with time-shift techniques. In specific, holographic techniques do also did not need the a priori knowledge of the propagation losses in the scan region and it also did not use any adjustable parameters such as the signal integration window to form an image. Nevertheless, the use of holographic techniques for BMR image formation has only been experimentally validated for monostatic scan protocols.

The goal of this paper is to determine the experimental feasibility of multistatic circular holography techniques for BMR image reconstruction. This works expands the preliminary results presented by the authors in a previous study that used only simulated data 15. The datasets used in this study were recorded from synthetic breast phantoms whose dielectric properties mimic the average permittivity and conductivity values of healthy and malignant breast tissues. The performance of the multistatic holographic reconstruction approach was assessed by calculating the SCR, contrast levels and spatial accuracy of the reconstructed images. Additionally, monostatic datasets from the same phantoms were collected and reconstructed using the holographic approach presented by the authors in a previous study14.  The resulting monostatic images were compared with their multistatic counterparts to assess the performance differences between both imaging protocols when holographic reconstruction techniques are used.

METHODS AND MATERIALS

II. A. Circular multistatic holography

Let consider given number of N transmitting and *M* receiving scan locations arranged in a circular geometry with radius *R*. If a waveform with a bandwidth B is radiated into the scan region from the scan location at , the collected responses on the receiving scan location at are given by:

(1)

where *T* is the total number of scattering objects in the scan area, , , are the polar coordinates of the qth scatter and  is its reflectivity . This is a generic circular radar signal model, where monostatic measurements are done in the case that . A diagram of the scan geometry can be seen in figure 1.

The signal model presented in (1) is given in the polar domain. As it can be seen in a sample dataset in figure 1, this representation makes it difficult to determine the location and nature of the breast structures in the scan area. In order to properly visualize the recorded responses from the breast structures, the dataset must be transferred to a rectangular system of coordinates. Microwave holography techniques achieve this by processing the spectrum of the recorded datasets. By applying a series of matched filters and frequency mapping processes, holographic techniques eliminate the phase wrappings introduced by the scan geometry and migrate the recorded BMR responses from the polar domain where they originally recorded to a Cartesian system of coordinates where it is easier to interpret them

The first step when using holographic reconstruction approaches is to calculate spectrum of the recorded responses. This is achieved by first calculating the Fourier transform with respect to the signal travel time, *t*, which is given by the following equation:

(2)

where and is the spectrum of. As you can notice, (2) is an amplitude-phase modulated function where the modulation factor is directly related to both the illumination and recording locations. To evaluate the effect of the phase wrappings caused by the scan geometry and the ones caused by the breast structure responses, the Fourier transform of (2) will be calculated with respect to and . Nevertheless, this is not a simple process due to the nature of the signalSoumekh. In order to calculate this spectrum, the stationary phase method is usedPapoulis. This technique evaluates the behavior of the instantaneous frequency of the signal to determine its spectrum. Using this approach, the Fourier transform of (2) with respect to and is given by:

(3)

where:

(4)

(5)

(6)

(7)

and are the amplitude components of the signal in the frequency space.

Equation (3) is also known as the spherical phase function of the system. Notice how terms (4) and (5) are a function of and , while (6) and (7) are only a function of *R*, which means that these phase wrappings are not related to the scatter objects inside the scan region. To eliminate these components from the spectrum of the recorded data, the following filter is used:

(8)

And by performing the following operation

(9)

Figure 2 shows the effect of this matched filter operation on the dataset previously shown in figure 1. Notice how the origin of the dataset is shift from the scan geometry to the center of the scan location.

Next, needs to be transferred to the frequency space which is associated with the rectangular system of coordinates related to the scanned region. To determine the nature of the mapping process, the inverse Fourier transform of (9) is calculated along the and directions. The resulting spectrum is given by:

(10)

where is the compensated magnitude of the spectrum. Notice how the term is not related to the angular location of the target. In order to make the phase modulation term of (10) a function of the target location, the following operation must be performed to (10):

(11)

Next, the spectral responses from must be transferred to . In order to do so, the following change of variable is used:

(12)

The trajectories in the plane are also known as iso-doppler contours. These lines in the signal multi dimensional spectrum correspond to locations that have the same frequency modulation indexSoumekh. Since and are defined such as:

(12)

(13)

The number of possible values in would be equal to *N*+*M*+1.

If we perform an integration of the spectrum responses along each possible value of in , the resulting spectrum would be given by:

(13)

Since is in a polar system of coordinates, a mapping process is required to transfer its information to the frequency space. This is achieved by using the following mapping functions:

(14)

(15)

The output of this mapping process is denoted as . Due to the nature of the mapping procedure described in (14) and (15), the sample locations on are not equally spaced. To process this dataset using standard Fourier techniques, an interpolation process is performed. Finally, to generate the reflectivity image of the scan region, the inverse 2D Fourier transform of is calculated yielding the final image.

II. B. Data acquisition Setup

The BMR datasets used in this study were recorded using a data acquisition system formed by a 56 × 56 × 40 cm plexiglass tank with two horn antennas attached to its support lid. The antennas in the system were placed with an angular separation of 135o to minimize the antenna crosstalk patterns. In all experiments, a multistatic radar configuration was used in which both S11 and S21 measurements were recorded using a Fieldfox N9923A vector network analyzer, which functioned both as the microwave waveform generator and recording device. The UWB waveform was irradiated and recorded using a horn antenna design developed by the authors, which was previously described in Saeed. Canola oil was used as matching medium.

The phantom materials using in this study mimic the actual dielectric properties of the breast tissues to emulate propagation losses and phase changes that affect the signal as it propagates in the breast region 4,7. The phantom was created using a styrene-acrylonitril cylinder with a diameter of 13 cm and a height of 35 cm. Breast tissue was simulated using glycerin, since its dielectric permittivity mimics the average values found in low density breast regions 12. The skin layer was created by covering this cylinder with a 1 mm layer of a mixture formed by a synthetic polymer, TX-151G, with distilled water in a ratio of 3 parts of water per part of synthetic polymer. Tumors were simulated using capsules filled with an 85% saline and 15% fructose solution. The dielectric characteristics of this mixture are described in detail in a study performed by Shinyashiki et al. 22. All the tumors used in the experiments presented in this paper had a spherical shape. Finally, the synthetic fibroglandular structures were generated using a mixture of 25% synthetic polymer compound and 75% distilled water. 23. These materials had similar dielectric properties to the values reported in various studies 8, 24, and setups with similar dielectric contrast have been used by several groups performing BMR research7, 17, 21, 23. The synthetic breast structures were attached to acrylic rods that connected them to a support structure. To evaluate the effect of the scattering generated by these acrylic rods, the authors collected 15 datasets with the rods at different locations within the phantom. In all scenarios, the scattering generated by the acrylic rods could not be distinguished from the background noise level. The dielectric properties of the phantom materials are given in Table I. A picture of the data acquisition setup is shown in figure 4.

II. C. Data acquisition protocol

The datasets were collected along a circular scan geometry with a radius of 14 cm. The phantom was positioned in the center of the system. The synthetic fibroglandular structures are attached to acrylic rods which were connected to a support base. This support based was attached via a nylon screw to the shaft of a stepper motor . This motor is used to rotate the antenna to a discrete set of locations in order to emulate a circular scan geometry. In this study, all datasets were collected using 144 different angular locations, which emulates 144 different scan locations. After the phantom contents are rotated into a particular angular position, a stepped frequency continuous wave with a bandwidth of 5.98 GHz and a center frequency of 3.01 GHz was used. Each waveform had 601 frequency points. The energy of each frequency component in the irradiated wave was 1mW. The system was calibrated using the same procedure as inFlores. In order to satisfy both the monostatic and bistatic angular sampling constraints, the spatial sampling along the  and scan directions satisfy the criterion formulated in Flores et al. 14.

(16)

where  is the radius of the scan region and is the wavelength corresponding to the maximum frequency component in the irradiated waveform. The spatial resolution along the signal travel direction inside the phantom is given by:

(17)

where *B* is the bandwidth of the irradiated signal.

The recorded data sets were processed using a conventional PC with 8GB of RAM. In all experiments a spherical capsule with 8mm diameter filled with the fructose and saline mixture was used to emulate a tumor. This size was chosen to simulate the dimensions of a non palpable lesion 25,26. The inclusion had a 3D nature to recreate the detection constraints that the algorithm would have to face on a more realistic scenario. In all cases, the targets were located at the same height as the antenna center.

II. D. Performance metrics

The performance of the proposed approach was quantitatively assessed by calculating the spatial accuracy, Signal to Clutter Ratio (SCR), Contrast to Clutter Ratio(CCR) and Tumor to Fibroglandular Response Ratio(TFRR) of the reconstructed images. The SCR was defined as:

(9)

where  is the maximum magnitude of the tumor signature and  is the standard deviation of the background. It was calculated as follows:

(10)

where  is the magnitude of the maximum magnitude of the fibroglandular structures signatures. The reconstructed images have a zero padding factor of two to improve its display. The second metric is the contrast to noise ratio, which is defined as follows:

(11)

This definition was adopted because the energy of the reconstructed images is commonly used to assess the presence of any lesions 7, 20, 21, 23. In all the reconstructed images the center of the phantom was be denoted as the origin. The spatial error was calculated by subtracting the centroid of the reconstructed tumor signatures area, delimited by the -3dB points, from the location of the inclusion in the experimental setup. To better visualize the reflections from the phantom interior, the skin reflections were removed using the algorithm proposed by Flores et al.27, and its location was indicated using a red circle. The criteria to classify a reconstructed response as a tumor were as follows:

1. The magnitude of the reflections in a candidate area are above 1.2x10-5 mW, which is the peak response of a 3 cm and a height of 3.5 cm ovoidal fibroglandular region responses. These are the average dimensions for a typical dense tissue area in an ACR2 breast. This energy value was obtained by calculating the average peak value a set of 30 images containing a synthetic fibroglandular structures positioned at different locations inside the phantom.
2. The area of the area of the reflections above 1.2x10-5 mW is at least 5mm2, which would correspond to the dimensions of an object with dimensions roughly half the size of the spatial resolution of the imaging system. This criterion is supported by the study presented by Fear et al.15 , in which it was shown that a BMR system built using using commercially available microwave instrumentation was capable of detecting objects with dimensions of at least .

III. Results

In this study, all datasets were taken using a phantom configuration in which two 3.5cm spherical dense tissue structures were positioned at (3,0) and (0,-3) cm. Given the bandwidth of the radiated waveform, the angular separation between scan locations and the size of the scan region, the responses from both dense tissue structures should be abled to be individually resolved in the reconstructed image. This size was determined by segmenting and measuring the fibroglandular structures in a set of 10 Magnetic Resonance Imaging datasets of patients at least 50 years old with an ACR2 breast density. In datasets where a tumor is present, the synthetic lesion was attached to a dense tissue structure located at five different experimental setups and their corresponding multistatic and monostatic reconstructed images are shown in this paper.

The first experimental setup and its corresponding monostatic and multiststic images can be seen in figure 4 a),b) and c) respectively. In this experimental setup, no tumor was present in the phantom. This dataset was recorded and reconstructed as a control dataset. It provides a guideline to visualize the responses of the phantom setup when no tumor is present. Notice how both dense tissue structures have very similar magnitudes. The second experiment setup is shown in figure 5a). In this scenario, a single tumor is located at the anterior location of the synthetic fibroglandular patch at (0,3) cm. The reconstructed monostatic and multistatic images can be seen in figure 5 b) and c). The multistatic image shows a more noticeable tumor signature compared with its monostatic counterpart.

In the third experimental setup, a synthetic tumor was attached at the dextral wall of the dense tissue structure at (3,0) cm. A diagram of the experimental setup can be seen in figure 6 a). The reconstructed monostatic and mulstistatic images can be seen in figure 6b) and c). The reconstructed images for this dataset show little difference, as the tumor signature location and magnitude with respect to the dense tissue structure is quite similar in both of them. This is a direct result of strong monostatic response caused by the direct illumination of the synthetic tumor. The fouth experimental setup and its corresponding monostatic and multiststic images can be seen in figure 7 a),b) and c) respectively. In this setup, the synthetic lesion was attached to the posterior part of the dense tissue patch at (3,0)cm. Similarly to the datsets reconstructed from experiment 2, the tumor signature in the multistatic image presents a higher contrast with respect to the dense tissue responses when compared with the lesion response in the monostatic image. Additionally, the location of the synthetic tumor in the multistatic image is closer to its location in the experimental setup than in the monostatic image.

Finally, the fifth experimental setup and the monostatic and multistatic images associated with it are shown in figure 8. In this scenario, the tumor is located in the sinistral side of the dense tissue structure at (3,0)cm. From the reconstructed images it can be appreciated that the tumor signatures have a lower magnitude than the reconstructed dense tissue structures. This is cause by the fact that the synthetic fibroglandular is reflecting most the irradiated waveform energy when the illuminating antenna is closer to the tumor. In order to provide a basis to perform a quantitative comparison between the monostatic and multistatic reconstruction approaches, the image performance metrics for each experiment are shown in Table II.

IV. DISCUSSION.

The research goal of this study is to assess the experimental feasibility of multistatic radar holography to reconstruct accurate BMR images. This approach has the potential of forming high contrast images without artifacts, while requiring less prior knowledge about the breast region compared to current time domain techniques. The results in this study are consistent with previous work performed by the authors, where preliminary results show that

The ultimate goal of this study was to assess the performance of circular holographic techniques in scenarios that mimic the dielectric properties of breast tissues. The potential of this approach lies on its capabilities to produce images without artifacts and no prior knowledge about the losses in the scan region. Previous work performed by the authors has shown that circular holography techniques were able to produce accurate images with higher SCR values than the current time-shift BMR reconstruction approaches. Nevertheless, multistatic holography required to be tested using phantoms that mimic the signal propagation losses present in BMR scenarios in order to assess its feasibility.

The authors decided to not perform a perfect impedance match between the immersion medium in which the antenna was immersed and the materials used to emulate the fatty breast tissues. This was done since in the majority of the BMR scan scenarios only an approximate range of the fatty breast tissue dielectric properties is known a priori and a perfect matching is quite difficult to achieve. In all the experimental setups, images reconstructed from multistatic datasets exhibited higher SCR, TFFR and CCR values compared to images formed using monostatic holography. Both monostatic and multistatic images presented similar spatial error values, which were smaller than the spatial resolution of the system(1cm). The proposed approach exhibited a short execution time, as the average time to reconstruct an image was 7.5 seconds. Although the execution time is 3 times longer than what it takes to form a monostatic image, 2.5 seconds, the processing time of the proposed approach is still 2 orders of magnitude lower than the time that it takes to collect a dataset (4 minutes).

The effect of dense tissue structures in the reconstructed BMR datasets was quantified and analyzed. The inclusion of fibroglandular regions was fundamental in order to assess the potential of the proposed approach to distinguish between dense tissue structure responses and tumor signatures. This requirement arises from the fact that in 80% of the cases breast tumors originate next to dense tissue structures. Since these structures usually are much larger than tumors and present a reduced contrast contrast, 2-1.1 to 1, with malignant lesions compared to fatty tissues(10-5:1), their responses can make difficult to the visualize the tumor signatures in the reconstructed images. From the control experiment, it can be seen that in scenarios where the tumor is not covered by the dense tissue structure, the energy of the tumor responses is between 40%-100% larger than the energy from the dense tissue structures responses.

The results of experiments 2 and 4 show the potential of the proposed approach when compared with the monostatic approach. In the images formed using monostatic holography, the tumor response has a magnitude of XXX mW, which is just a fraction of the magnitude of the tumor signature in the multistatic approach. The images formed from the monostatic datasets corresponding to these experiments the are consistent with the results presented by the authors in a previous study14. This study showed that tumor signatures in images formed from experimental datasets where the tumor is located at the anterior or posterior locations of a dense tissue structure had a similar magnitude than the adjacent fibroglandular region. The higher magnitude of the tumor responses in the multistatic images results in better SCR, CCR and TFFR values for the multistatic images. Notice also that the tumor responses have a higher magnitude than the adjacent dense tissue structures. Additionally, the images formed from the multistatic datasets show a smaller spatial error and a more defined tumor signature.

When the results of experiments 2 and 4 are compared with experiment 3 it can be noticed that there is a decrease in the performance of both multistatic and monostatic approaches. The reduction of SCR and CCR in experiments is in the order of 50% compared to experiment number 3. This is caused by the reduced illumination caused by the position of the tumor with respect to the dense tissue structure. In experiment 3 the tumor is at the dextral side of the dense tissue structure, resulting in an increased number of locations from where the tumor can be directly illuminated. More importantly, the tumor is directly illuminated by the antenna when the distance between them is at its minimum, which is not the case in experiments 2 and 4. Nevertheless, in all the multistatic images the TFFR are above 3dB, which means that the reconstructed tumor signatures have a magnitude that is at least 50% higher than the dense tissue responses. Additionally, the SCR and CCR are above 4dB, showing that the tumor signatures and the differences between the lesion and the dense tissue structure responses are considerably above the clutter levels in the image.

The results from experiment 5 show that lesions are difficult to detect in the image in scenarios where they are located in a position where most of the radiated UWB waveform is reflected by a nearby dense tissue structure. This is caused by the high dielectric contrast between the dense tissue and the fatty tissues, which result in a large impedance leading the reflection of a considerable portion of the incoming wave. Additionally, the portion of the wave that penetrates into the dense tissue structure is affected by a large attenuation factor due to the high conductivity of the material used to emulate the fibroglandular structures. Although in all experiments the SCR is within the same order of magnitude as in experiments 2,3 and 4, the CNR and TFFR are under 3dB, making difficult to detect if that response is caused by a tumor or a sidelobe of the dense tissue structure response. This is one of the most challenging imaging scenario for most of the current AMI techniques. Nevertheless, recent studiesstasi have shown that when BMR is used in conjuction with MT tumors in this type of scenarios can be successfully detected.

The SCR of the images corresponding to experiments 2,3 and 4 was in average 15 dB, which is 100% higher than the value (5.1dB) obtained in the most recent multistatic experimental studies that included both synthetic tumors and fibroglandular structures on its phantoms 21, in which time domain techniques where used to form the images. There is no comparison point for the images formed from experiments 1 and 5, as to the best of the authors knowledge no experimental study has been pubilished in which the responses from a phantom containing only dense tissue structures, or a phantom where the tumor is obstructed by a dense tissue structure has been published. The improvement on the performance of the multistatic approach is the used of the multidimensional phase behavior of the recorded responses to identify reflections from different sources at different scan locations in scenarios with multiple scattering objects and coherently combine them and transfer them to their location in the scan region. As shown in previous studies performed by the authors, time-shift techniques present limitations to reconstruct datasets containing multiple inclusions, yielding images with lower SCR values than the ones obtained using holographic approaches 14.

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