Time-of-Arrival Calibration for Improving the Microwave Breast Cancer Imaging

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Abstract — In radar based confocal microwave imaging for breast cancer detection, recorded data are synthetically focused to a confocal point within the breast. This is the basis for both data-independent and data adaptive methods to form the breast image and can be enhanced by multistatic approach. This approach inherently assumes that the propagation velocity depends only on the average dielectric property of the breast. However, in real cases, the breast tissues are inhomogeneous and therefore the propagation velocities vary for different propagation paths. Thus, use of an average propagation velocity can result in false localization. This paper proposes an auto-calibration method to compensate the time-of-arrival from confocal point to the receiving antennas. We demonstrate using simulations on FDTD numerical breast phantoms that the proposed method helps to form an enhanced image for inhomogeneous breast.

Index Terms — TOA, Breast Cancer, CMI, Beamforming, FDTD.

I. INTRODUCTION

Breast cancer is among the most common cancers affecting women’s health all over the world. Early detection is the best protection in terms of successful treatment and long-term survival. Radar based confocal microwave imaging (CMI) technique has shown some potential for the detection of early stage cancers [1]. In addition, it is non-ionizing, non-invasive, and can be sensitive to small-sized tumors.

In CMI technique, a low power ultra wideband pulse is radiated by antennas in to the breast. A significant portion of the incident UWB electromagnetic waves will be reflected by the breast skin. A small portion of incident waves hopefully will penetrate through the skin and travel through the breast to interact with the malignant and other benign and healthy breast tissues. The interaction with these tissues would result in scattered energy in every direction. The receiving antennas are employed outside the skin to capture the scattered energy. The strength of scattered energy available at the receivers depends on the complex microwave dielectric properties of malignant, benign and other breast tissues. The essence behind the confocal microwave imaging is the available dielectric property contrast between the malignant and other breast tissues. Early research on this topic identified the dielectric contrast between normal and malignant tissues to be as high as 5:1 [1]. However, recent experimental investigations using surgical data revealed that while dielectric contrast between fatty and malignant tissues could be as large as 10:1, the contrast between glandular to malignant tissues can be as low as 1.2:1 [2]. Such a low dielectric contrast can pose real challenges to the detection of malignant tissues using any of the radar based microwave imaging techniques [3].

Confocal microwave imaging (CMI) initially used simple Delay-And-Sum (DAS) beam former to enhance the tumor response from the received scattered energy [1]. Later, a data adaptive method known as MAMI was proposed which uses robust capon beamformer (RCB) to improve the image resolution [4]. Recently, a series of published works have experimentally demonstrated the CMI method on breast phantoms [5], [6]. However, all these existing methods ideally align all the calibrated data to a synthetic confocal point before calculating the energy. They also assume that all the propagation paths between breast tissues and the receivers have the same propagation velocity so that the time of arrival (TOA) from a confocal point to a receiving antenna depend only on the round-trip distance. In fact, the breast heterogeneity makes the propagation velocity vary for each of the propagation path taken by the scattered signals. Some researchers have proposed estimating the TOA using an earlier estimate of Tumor’s position to iteratively reduce the error [7].
However, the low dielectric contrast that exists between benign (glandular) and malignant tissues poses severe challenges that warrant accurate calibration to achieve effective discrimination of the tissues. With this aim, in this paper we propose an auto-calibration method using TOA to compensate for the effects of tissue heterogeneities. The proposed auto-calibration approach will enhance the image reconstruction algorithm by making it sensitive to the patient-dependent breast tissue variability.

II. BREAST PHANTOMS

We employ FDTD based numerical 2-D and 3-D breast phantoms that use the more accurate two-pole Debye models to cover UWB frequency band from 0.5-20GHz. The models fit the latest surgical data very well and are easy to incorporate with the FDTD models. The two-pole Debye model is defined by [8]:

\[
\varepsilon(\omega) = \varepsilon_\infty + \sum_{n=1}^{N} \frac{\Delta \varepsilon_n}{1 + j \omega \tau_n} + \frac{\sigma_s}{j \omega \varepsilon_0}
\]

Equation (1) could obtain dielectric properties for the malignant as well as high adipose (mostly fatty but less of fibroconnective or glandular), medium adipose, and low adipose (less of fatty but mostly fibroconnective and glandular) breast tissues. The Debye parameters for those three tissue categories are given in Table I.

We first employ a 2-D FDTD breast phantom as illustrated in Fig. 1 for simulations. A semi-circular shaped breast model with a radius of 50mm is used along with the chest wall. The skin layer is assumed to be 2mm thick. A circular arc antenna array with 22 ideal antenna elements is utilized and is located 10mm away from the skin. All the three main categories of the breast tissues are simulated in terms of their dielectric properties. As shown in Fig.1, the malignant tissue is surrounded by many medium adipose and glandular tissues whose dielectric properties are very close to that of malignant tissues. For the purposes of TOA estimation, the malignant tissue is modeled as having a circular shape with a diameter of 10mm. An incident modulated Gaussian pulse with frequencies up to 7.4 GHz is used to illuminate the breast. We also used a 3-D breast model as shown in Fig. 2 in which the fatty, medium adipose, glandular tissues are scattered within the breast. The malignant tissue is located in the lower centre with co-ordinates (15, 0, 25) mm and a concentric circular antenna array with 44 ideal antenna elements (blue dots in Fig. 2) is placed 10mm away from the skin. Chest wall and the surrounding skin are also included. The skin layer is 2mm thick and the first-order Debye model is used for skin (for 2D and 3D models): \( \varepsilon_\infty = 18.4, \Delta \varepsilon = 21.9, \tau = 17.5\text{ps} \) and \( \sigma_s = 0.737 \text{S/m} \). The chest wall is also included: \( \varepsilon_\infty = 6.75, \Delta \varepsilon = 47.91, \tau = 10.1\text{ps} \) and \( \sigma_s = 0.85 \text{S/m} \).

III. AUTO-CALIBRATION FOR TOA

The breast is inhomogeneous as shown in Figs. 1 and 2. Hence, the propagation velocities can be different for signals reaching at different receiving antennas resulting in different TOAs. Thus, the differences in TOAs must reflect during the process of aligning the data to a confocal point. To achieve this, firstly, the initial TOA for any propagation path need to be determined which can be done using any of the existing techniques [1], [3]-[6]. However, in the next step, one needs to focus the early-time content to a known reference antenna position using a beamformer. Here, we took the centrally located antenna as the reference position using DAS beamformer. We modify the initial TOA by deriving a compensation factor that minimizes the variance of the focused energy. The compensation factor varies until the estimated focusing point coincides with the reference antenna position. This

<table>
<thead>
<tr>
<th></th>
<th>High Adipose</th>
<th>Medium Adipose</th>
<th>Low Adipose</th>
<th>Malignant</th>
</tr>
</thead>
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<tr>
<td>( \Delta \varepsilon_1 )</td>
<td>0.58</td>
<td>19.64</td>
<td>20.81</td>
<td>25.61</td>
</tr>
<tr>
<td>( \Delta \varepsilon_2 )</td>
<td>1.09</td>
<td>14.23</td>
<td>20.22</td>
<td>23.91</td>
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<tr>
<td>( \tau_1 \text{ ps} )</td>
<td>8.07</td>
<td>5.81</td>
<td>7.39</td>
<td>7.22</td>
</tr>
<tr>
<td>( \tau_2 \text{ ps} )</td>
<td>19.25</td>
<td>16.49</td>
<td>15.18</td>
<td>15.30</td>
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<tr>
<td>( \varepsilon_\infty )</td>
<td>3.14</td>
<td>5.57</td>
<td>7.82</td>
<td>6.75</td>
</tr>
<tr>
<td>( \sigma_s \text{ S/m} )</td>
<td>0.036</td>
<td>0.52</td>
<td>0.71</td>
<td>0.79</td>
</tr>
</tbody>
</table>
compensation factor is used to calibrate the TOA. The effects of this process can be observed in Fig. 3 in which the data from 3D model (as shown in Fig. 2) is used. We chose the reference antenna at (0, 0, 60) mm as the transmitter and the other 43 antennas to be receivers. For different compensation factors, the variation of the focused position is shown in Fig. 3. When the position is focused to the correct position, that compensation factor is selected as indicated in Fig 3(c). The effects of the different compensation factors are discussed in the next section.

Thus, our proposed auto-calibration method incorporates the variation in TOAs for different propagation paths for it to align the pre-processed data properly to the confocal point without any ambiguity. This is a crucial step for the reconstruction algorithms to form an image based on scattered energy which assumes significance in the face of reduced dielectric contrast.

IV. SIMULATION RESULTS

To verify the proposed method, we applied it on both 2D and 3D computational FDTD models. In our first example, the data from our 2D model (Fig. 1) is processed. The results shown in Fig. 4 (a) and 4(b) correspond to inaccurate tumor localizations due to differences in the estimation of time-of-arrival. When the time-of-arrival with proposed auto-calibration is used to compensate the bias caused due to the random breast tissue heterogeneity, the right tumor location is obtained as shown in Fig. 4(c). The compensation factors obtained for Fig 4(a)-4(c) are 0.8, 0.7, and 0.74, respectively. Our second example uses the data from the 3D FDTD model that is shown in Fig. 2. The results shown in Figs 5(a)-5(c) confirm that the tumor location will vary with different time-of-arrival estimation and the use of proposed calibration method result in the correct tumor location as shown in Fig. 5(c).

Fig. 3. Auto-calibration. (a) compensation factor=2.5. (b) compensation factor=2.9. (c) compensation factor=3.3.

Fig. 4. Breast images. (a) compensation factor =0.8. (b) compensation factor=0.7. (c) compensation factor=0.74.
V. CONCLUSIONS

We proposed a novel auto-calibration method which utilizes the antenna position in a prior knowledge to compensate the TOA estimation. As the incident signal is dominant in the early-time content, if we focus the early-time content, the focused position will indicate the incident antenna position. Only for accurate TOA, the estimated antenna position will coincide with the true antenna position. In this way, we can find out the proper TOAs for inhomogeneous breast.

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