Initial Study on In Vivo Conductivity Mapping of Breast Cancer Using MRI

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Purpose: To develop and apply a method to measure in vivo electrical conductivity values using magnetic resonance imaging (MRI) in subjects with breast cancer.

Materials and Methods: A recently developed technique named MREPT (MR electrical properties tomography) together with a novel coil combination process was used to quantify the conductivity values. The overall technique was validated using a phantom study. In addition, 90 subjects were imaged (50 subjects with previously biopsy-confirmed breast tumor and 40 normal subjects), which was approved by our institutional review board (IRB). A routine clinical protocol, specifically a \( T_2 \)-weighted FSE (fast spin echo) imaging data, was used for reconstruction of conductivity.

Results: By employing the coil combination, the relative error in the conductivity map was reduced from \( \pm 70\% \) to \( \pm 10\% \).

The average conductivity values in breast cancers regions (0.89 \( \pm \) 0.33 S/m) was higher compared to parenchymal tissue (0.43 S/m, \( P < 0.0001 \)) and fat (0.07 S/m, \( P < 0.00005 \)) regions. Malignant cases (0.89 S/m, \( n = 30 \)) showed increased conductivity compared to benign cases (0.56 S/m, \( n = 5 \)) (\( P < 0.05 \)). In addition, invasive cancers (0.96 S/m) showed higher mean conductivity compared to in situ cancers (0.57 S/m) (\( P < 0.0005 \)).

Conclusion: This study shows that conductivity mapping of breast cancers is feasible using a noninvasive in vivo MREPT technique combined with a coil combination process. The method may provide a tool in the MR diagnosis of breast cancer.

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The dielectric properties of breast tissue, especially its electrical conductivity (denoted as \( \sigma \) in units of Siemens/meter (S/m)), have long been investigated in vitro and in vivo using electrical impedance tomography (EIT) systems.1–7 Findings have consistently shown that malignant breast cancers present elevated conductivity values when compared to normal tissue. The conductivities of malignant mammary samples are typically in the range of 0.8 ~ 1.4 S/m, while those of normal tissue are between 0.1 ~ 0.2 S/m at a frequency of \( \sim 100 \) MHz.8–5 The biological mechanism behind the elevated conductivity is attributed to factors such as the presence of necrosis and cell membranes breakdown, increased cell membrane charge, sodium concentration, and water content.8 These prior published results were produced from in vitro samples or through the use of EIT systems with low spatial resolution. The ability to study conductivity in a noninvasive manner using in vivo magnetic resonance imaging (MRI) would potentially have great merit.

The capability of MRI to probe the conductivity of tissue in a noninvasive manner was first demonstrated over 20 years ago.9 Since then, relatively few studies have reported on the use of the technique.10 However, MR-based electrical property imaging, commonly referred to as magnetic resonance electric property tomography (MREPT), has recently been the subject of active investigation mainly due to the advent of high-field systems.11–16 While several initial studies have demonstrated the potential use of monitoring electrical properties in vivo for clinical purposes, the effectiveness of the technique has yet to be determined.17–20

Technically, measurements of the electrical conductivity using conventional MR imagers require knowledge of radio frequency (RF) field distribution, ie, the \( B_1 \) map. Especially, the electrical conductivity maps can be reconstructed using only the phase information of RF field.12 However, this phase-based reconstruction is valid when the magnitude of transmit field (denoted \( B_{1m} \)) and receive...
field (denoted \(B_{1m}\)) is homogeneous. Hence, we developed a method to improve the validity of the phase-based reconstruction and we apply this technique to investigate the conductivities from normal and patients with breast cancer including noninvasive, invasive, and malignant cancers.

**Materials and Methods**

**RF Field Distribution (\(B_1\) magnitude and Phase Information)**

The complex valued \(B_1\) field can be decomposed into two circularly polarized fields that contribute to signal formation in MRI \(^{21}\); one field rotates in the clockwise direction (\(B_1^-\) by convention) while the other rotates in the anticlockwise direction (\(B_1^+\) by convention). The magnitude of \(B_1^\pm\) is related to the transmit coil sensitivity and interacts with the magnetization for spin excitation, whereas the magnitude of \(B_1^\pm\) is related to the receive coil sensitivity. While \(B_1^-\) and \(B_1^+\) have phase attributes, ie, they are complex valued, the conventional “\(B_1\) mapping” techniques often employed in clinical MRI scanners involve measuring \(B_{1m}\) information.\(^{22}\) While the absolute phase of \(B_1^-\) is difficult to ascertain, its value can be estimated under certain assumptions by taking half the value of the phase from a spin echo image.\(^{10,16,23}\) In contrast, \(B_1^+\) cannot be measured directly and the unavailability of \(B_1^-\) can be problematic for electrical conductivity mapping if large spatial variations of \(B_{1m}\) are present, as will be shown.\(^{24}\) The technical challenge addressed here is to reduce the spatial variations of \(B_{1m}\) and improve conductivity quantification.

**Conductivity Reconstruction Using Phase-Based EPT**

For full determination of the conductivity, the time-harmonic Maxwell equation must be solved with measured \(B_1\) values. Under the assumption of locally constant dielectric values, it can be reduced to the form in Appendix A Eq. A1. Furthermore, it has been shown that the conductivity is primarily determined by the phase of the \(B_1\) field.\(^{12}\) Using the transceive phase \((\phi_{\pm})\), which can be easily obtained by taking the phase value of a typical spin echo-based MR sequence,\(^{16}\) the conductivity map can be reconstructed as:

\[
\sigma \approx \frac{\nabla^2 \phi_+}{2\mu_0\omega} \quad (1)
\]

This method is referred to as phase-based EPT,\(^{12}\) since only the phase of \(B_1\) is used. The approximation of Eq. 1 is valid with the assumption that the spatial variation of \(B_{1m}^-\) and \(B_{1m}^+\) is negligible.\(^{12}\) Invalidity of Eq. 1 generates artifact which is presented as a spatially varying conductivity value for a homogeneous region. Details regarding the derivation of Eq. 1 are included in Appendix A.

**Data Combination for Conductivity Reconstruction**

The assumption leading to Eq. 1 holds well when using a quadrature body coil (QBC) or single-channel homogeneous transmit-receive coil.\(^{12,13,25}\) When a receive coil with spatially varying sensitivity is employed, such as in the case of using multichannel receive coils, directly using the phase-based EPT is not appropriate. For example, in typical breast imaging, multichannel receive coils with varying sensitivities are now commonly used and, thus, conductivity reconstruction following a simple summation of the individual coil phase data or an averaged conductivity image following reconstruction from individual coils will lead to error. To overcome this problem, we propose an alternative data combination of multicoil data. Here the signals at each of the \(j\) \((j=1,\ldots,N)\) where \(N\) is the number of receive coils) were summed after zero-order phasing each \(j\)-th coil \((\phi_{0j})\) so as to minimize the spatially varying sensitivity term (see Appendix B for details). Ideally, the zero-order phase values should be obtained per subject. However, this is time-consuming and it is difficult to generate an image without the contrast information required (Eq. B5). Here, these zero-order phasing values were obtained from a separate reference scan conducted using a homogeneous phantom. Phantoms with known conductivity values were placed inside the multichannel receive coil and scanned. The zero-order phase for each coil was determined via the aforementioned method. To illustrate the conductivity reconstruction performance, the proposed method and a method which did not perform any phasing was compared both visually and using a relative error of the resulting conductivity map \((\Delta)\) calculated as:

\[
\Delta(\%) = \left| \frac{\sigma_{\text{true}} - \sigma_{\text{est}}}{\sigma_{\text{true}}} \right| \times 100(\%)
\]

where \(\sigma_{\text{true}}\) is the true conductivity value measured by using a conductivity meter (HI8733N, Hanna Instruments, Woonsocket, RI) and \(\sigma_{\text{est}}\) is the conductivity estimate measured using MR-based conductivity mapping technique.

**Reconstruction Process**

The entire reconstruction process for breast conductivity estimation is summarized in Fig. 1. In the initial setup stage, fast spin echo (FSE) data from a reference phantom were acquired to determine the optimal signal combining phase values \(\phi_{0j}\) of the breast receive coil. Specifically, the continuous \(\phi_{0j}\) value was discretized to 16 steps and then a discrete optimization procedure was performed to solve Eq. B5. After acquiring the \(\phi_{0j}\) values from the reference phantom, the values were employed to combine the in vivo FSE data. As mentioned, the reference phantom study needs to be conducted only once and the phase values acquired from phantom study were used in all subsequent in vivo experiments.

After the data combination process, the conductivity of breast tissue was reconstructed with the newly combined phase image. A weighted polynomial fitting routine was employed to calculate \(r\).\(^{26}\) This method is applied to reduce boundary artifact and statistical noise in \(B_1\) information.\(^{17,27}\) Conductivity images of the subjects were produced from the calculated values after bilateral (kernel size = 17) and median filtering (kernel size = 2). For weighted polynomial fitting and bilateral filtering, a weighting factor, \(w\), based on the intensity dissimilarity, was generated using an FSE magnitude image (Fig. 1).\(^{28}\) All postprocessing was conducted on a PC (Intel Pentium Processor 2.4 GHz) using MatLab R2011a (MathWorks, Natick, MA). After all postprocessing, artifact in conductivity map was examined (by J.W.S. and S.Y.K.) by visually comparing the FSE images with the conductivity maps.

The average conductivity values from the tumor and normal regions were compared using ROI (region of interest)
determined from the dynamic and FSE images. Regions with hyperintensity in the FSE image were selected as fat. The ROI of breast cancer was selected by a board-certificated radiologist with 12 years of expertise in breast MRI. From the region excluding fat, regions with hyperintensity in the dynamic contrast-enhanced (DCE) image were selected as cancer and the remaining regions were assigned as the parenchyma region. To investigate whether invasiveness of tumor can be differentiated, statistical analysis was performed between invasive and in situ cancer (ductal carcinoma in situ) regions using Student’s $t$-test. Additional tests were conducted to determine whether there is any dependence of conductivity as a function of cancer size or age.

**In Vivo Breast MRI**

Breast studies were performed in a 3T clinical scanner (MR750, GE Healthcare, Waukesha, WI) with an 8-channel breast receive coil. MRI data from a total of 90 female patients were obtained. The subjects ranged in age from 34 to 74, with a mean age of 52.3 years ± 11.6. Among the 90 subjects, 40 did not have breast tumor and 50 had breast tumor recently confirmed by imaging-guided biopsy. All 50 patients did not yet have neoadjuvant chemotherapy or surgery for breast cancer when they were imaged on breast MRI. Among 50 patients with breast tumor, 15 patients with breast cancer smaller than 10 mm were excluded due to the limited spatial resolution required for reconstruction. Among 35 patients, we reconstructed conductivity for 23 invasive cases, seven ductal carcinoma in situ (DCIS) cases, and five benign lesions (fibroadenoma in four, fibroadenomatoid hyperplasia in one) that were confirmed by ultrasonography-guided biopsy and that were seen on MRI. Therefore, we reconstructed a total of 75 cases (40 normal breast fat and parenchyma, 30 breast cancers, and five benign lesions, summarized in Table 1). Among 30 breast cancers, there were 19 luminal A, 2 luminal B subtypes, 5 HER-2, and 4 triple-negative subtypes. This study was approved by our Institutional Review Board (IRB). Informed consent was waived by the IRB since the study was retrospective and involved postprocessing of clinical data.

The examination protocol employed consisted of five imaging sequences: $T_1$ fast spoiled gradient echo (FSPGR), $T_2$ FSE, $T_2$ stimulated inversion recovery (STIR), diffusion-weighted imaging (DWI), and DCE-MRI. The kinetic DCE uptake curve was produced with a computer-aided detection system (CAD stream software, v. 5.2.8.591, Merge Healthcare, Milwaukee, WI).

**TABLE 1. Patient Data Summary for Those Processed ($n = 75$)**

<table>
<thead>
<tr>
<th></th>
<th>No. of cases</th>
<th>Age range</th>
<th>Cancer size (mm)</th>
<th>Intrinsic subtype</th>
</tr>
</thead>
<tbody>
<tr>
<td>Normal</td>
<td>40</td>
<td>52.3 ± 11.6</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Benign fibroadenoma</td>
<td>4</td>
<td>38 ± 13.5</td>
<td>17.2 ± 3.2</td>
<td></td>
</tr>
<tr>
<td></td>
<td>1</td>
<td>42</td>
<td>14.3</td>
<td></td>
</tr>
<tr>
<td>Malignant invasive ductal carcinoma</td>
<td>18</td>
<td>51.8 ± 10.3</td>
<td>21.0 ± 7.9</td>
<td>Luminal A ($n = 10$)</td>
</tr>
<tr>
<td></td>
<td>2</td>
<td>2</td>
<td>HER-2 ($n = 2$)</td>
<td></td>
</tr>
<tr>
<td></td>
<td>4</td>
<td>2</td>
<td>Triple negative ($n = 4$)</td>
<td></td>
</tr>
<tr>
<td>tubular carcinoma</td>
<td>3</td>
<td>52.0 ± 6.9</td>
<td>18.3 ± 7.4</td>
<td>Luminal A</td>
</tr>
<tr>
<td>mucinous carcinoma</td>
<td>1</td>
<td>58</td>
<td>30.2</td>
<td>Luminal A</td>
</tr>
<tr>
<td>lobular carcinoma</td>
<td>1</td>
<td>46</td>
<td>30</td>
<td>Luminal A</td>
</tr>
<tr>
<td>ductal carcinoma in situ</td>
<td>7</td>
<td>51.7 ± 10.0</td>
<td>20.8 ± 7.6</td>
<td>Luminal A ($n = 4$) HER-2 ($n = 3$)</td>
</tr>
</tbody>
</table>
The $T_2$-weighted FSE used the following parameters: field of view (FOV) = 320 mm, 50 slices (3 mm thickness and no gap), TR/C24 = 4420 msec, TE$_{eff}$ = 102 msec, ETL = 20, matrix size = 416 x 256, total imaging time ~2 min. Phase data of the FSE sequence were retrieved for conductivity reconstruction, hence no additional scans were necessary.

Results

Figure 2 shows the results of the phantom experiment. The magnitude, phase, and conductivity images obtained using a conventional data combination routine (complex sum without phasing) and our proposed data combination method is displayed in Fig. 2a,b, respectively. The resulting magnitude image shows a more homogeneous image intensity distribution after our data combination method (Fig. 2b). The pattern of the transceive phase map used for conductivity reconstruction was also altered. The resulting conductivity map (Fig. 2a,b) and its average quantitative value (Fig. 2c) are provided. When the conventional method for data combination was employed, the average conductivity values were 0.56 S/m (right) and 0.35 S/m (left). However, when the data were combined with the $\phi_{u,j}$ sets (proposed method), the average conductivity values changed to 2.03 S/m (right) and 0.89 S/m (left), which were close to the true values, 2.15 S/m (right) and 1.05 S/m (left). By employing the proposed method, the relative error of the resulting conductivity map ($\Delta$) also decreased to from ~70% to 10%. There still remained artifacts in the conductivity map (white arrow in Fig. 2b) due to the spatial variation of $B_{1m}$. However, this artifact was observed at the boundary between air and object where the $B_{1m}$ is rapidly varying.

Figure 3a shows the average and standard deviation (SD) of the conductivity values acquired from cancer, fat and parenchyma regions from all subjects. Average conductivity values of cancer regions (0.89 S/m) were higher than that of other normal breast tissue regions. For normal breast tissue including parenchyma and fat, there was no apparent difference in the conductivity values from patients and normal subjects. The average conductivity values ($\sigma_{fat}$: 0.07 S/m, $\sigma_{parenchyma}$: 0.43 S/m, $\sigma_{cancer}$: 0.89 S/m) measured in this study correspond well with the values ($\sigma_{fat}$: 0.04 S/m, $\sigma_{parenchyma}$: 0.2 S/m, $\sigma_{cancer}$: 0.8−1.4 S/m at 100MHz) measured at ex vivo.3–5 By biopsy, 30 cases were confirmed as malignant and five cases were confirmed as benign. Among the malignant cases, 23 cases were confirmed as invasive and seven cases were confirmed as ductal carcinoma in situ (Table 1). The conductivity values of cancer between malignant and benign were compared and are shown in Fig. 3b. The conductivity values of benign lesion were generally lower than those of malignant cancers ($P < 0.05$). However, three benign cases showed larger conductivity values than the lowest conductivity value of malignant cancers (0.46 S/m).

The conductivity values of cancer between invasive and in situ case are compared in Fig. 3b. The conductivity values of in situ cancers were significantly lower than those of invasive cancers ($P < 0.0005$). However, the dependence between cancer type and its conductivity seemed to range widely for invasive cases. The conductivity values were widely distributed from 0.55 to 1.65 S/m.

Figure 4 shows representative FSE images, dynamic images, conductivity images, ADC maps, and kinetic curves from five patients of different cancer types. The reconstructed conductivity maps reveal elevated values in regions identified as invasive cancer areas. In general, there were slight variations of conductivity values in the breast (white arrow). This is due to inhomogeneous $B_{1m}$ remaining after calibration. This also can be seen in the corresponding FSE image regions. This did not seem to cause problems in the overall calculation of the conductivity of cancer. In the images
provided, three invasive and one in situ cancer showed a rapid initial uptake of contrast material and washout at the delayed phase of enhancement on DCE images along with low ADC values (<1.6 × 10⁻³ mm²/s) on DWIs. One benign case (Fibroadenoma) showed a rapid initial uptake of contrast material and persistency at the delayed phase of enhancement on the DCE image along with high ADC values (2.61 × 10⁻³ mm²/s) on DWI. The ADC values seemed to have an inverse proportionality with the conductivity, which is in agreement with a recent animal study.

Finally, the relationship between conductivity values of cancers with other factors is analyzed in Fig. 5. There was negligible dependence of the conductivity with cancer size (Fig. 5a). The conductivity values of cancers were slightly dependent on the age of the patient but it did not seem to be a dominant factor (Fig. 5b, R² = 0.0396).

**Discussion**

In this study we demonstrated the feasibility of obtaining conductivity maps in vivo from patients with breast cancer using a data combination technique that can be readily incorporated into a standard clinical imager. Breast cancer regions gave elevated conductivity values when compared to normal tissue, which was in agreement with previous studies where in vitro samples and other modalities such as EIT were employed. Malignant cases (30 cases) showed...
higher conductivity than benign cases (five cases). In addition, invasive cancers showed higher conductivity compared to DCIS. As described in previous studies, conductivity of human tissue at the MR Larmor frequency (in this study, 128 MHz) mainly depends on ion concentration,4,8 water content,8,30 bound water portion,31,32 etc. Variation of these factors may result in differences of conductivity values between normal and cancerous tissue. It remains a question as to how much these factors influence in vivo conductivity values. Recently, an animal study for tumor showed the correlation between conductivity measured using an impedance analyzer and the ADC value29 and the result showed that conductivity was inversely proportional to the ADC value. The correlations between EPT with other physiological parameters such as ion concentration, diffusion, and perfusion also are worth investigation.

The benefits of using MRI as a tool for conductivity estimation are enhanced spatial resolution when compared to other modalities and the noninvasive nature of the technique. In addition, the imaging procedure employed to obtain conductivity maps was a routinely used clinical protocol. Therefore, neither additional scan times nor a new scanning sequence was required to generate the maps and the imaging parameters of the original clinical protocol need not be modified. The study does not require contrast agents, which can benefit noncontrast breast screening. The feasibility to differentiate between parenchyma and lesions shows promise to further investigate whether conductivity can be a marker for differentiating tumor malignancy and cancer aggressiveness, or whether conductivity has prognostic value. In this study, 10 patients with benign tumor were imaged but five cases were excluded due to the size of the lesion. Therefore, as of now it is hard to ensure the relationship between the conductivity of malignant and benign, although the preliminary results correspond to a previous study.33 With more benign cases, further study can be performed.

A novel data combination process from multichannel receive coil data was developed which increased the accuracy of conductivity value estimation. Without this process, the quantification step was prone to error. A reference phantom study was required to determine the phase correction for each individual coil. The procedure only needs to be conducted once as long as the same receiver coil is used. Here, a global phase set was employed due to its simplicity. There still remained artifacts in the conductivity map of patients and those were usually observed near the air–tissue interface. These are caused by the individually differing patterns of the $B_{1m}$ that is different from the reference phantom $B_{1m}$ pattern. For the data used in this study, this type of artifact did not affect the evaluation of the conductivity of breast cancer and parenchyma. If there is a need to enhance the quality of conductivity map, a subject-specific phase set should be individually acquired and used.

There was no effect of the conductivity estimates related to cancer size studied in this population. However, for cancers whose size was small (typically below 10 mm), it is hard to acquire an accurate conductivity value due to the limited spatial resolution and lack of signal-to-noise ratio (SNR). Equation 1 requires a minimum of 3 pixels for one-dimensional calculation and therefore we excluded those small-sized cancers. This is a limitation of the study, which used a conventional clinical scan time to acquire the data. The limitation should be more closely investigated to analyze the tradeoffs between cancer size and accuracy of conductivity estimation and determine whether alternative acquisition sequences such as a bSSFP (balanced steady-state free precession) which provides better SNR efficiency should be used. However, problems in bSSFP images such as banding artifacts, phase jumps near fat region should be considered. For cancers whose margin was irregular, it was also hard to determine the conductivity value due to heterogeneity inside the cancer region. It seems questionable what

FIGURE 5: Relationship between conductivity values of breast cancer as a function of (a) cancer size and (b) patient age.
value of conductivity is being measured from the complex structures.

In conclusion, this study shows that conductivity mapping of breast cancers is feasible using noninvasive in vivo MREPT technique combined with a coil combination process. By employing this method, the difference in conductivity values according to invasiveness and malignancy can be observed. This method may provide a tool in the MR diagnosis of breast cancer.

Appendix A: Phase-Based Conductivity Reconstruction

In MREPT, electrical conductivity in a constant piecewise region is reconstructed from RF magnetic field \( B \) information using the Helmholtz equation, which can be decomposed into its magnitude and phase components as:

\[
\sigma = \frac{1}{2\mu_0\Omega} \text{Im} \left\{ \frac{2 \nabla B_{1m}^+ \cdot \nabla e^{i\phi_+}}{B_{1m}^+ e^{i\phi_+}} + \frac{\nabla^2 e^{i\phi_+}}{e^{i\phi_+}} + 2 \frac{\nabla B_{1m}^- \cdot \nabla e^{i\phi_-}}{B_{1m}^- e^{i\phi_-}} + \frac{\nabla^2 e^{i\phi_-}}{e^{i\phi_-}} \right\}
\]

\[
(A1)
\]

which can be interpreted as the average conductivity value reconstructed from \( B_{1m}^+ \) and \( B_{1m}^- \) separately. When the spatial variations of \( B_{1m}^+ \) and \( B_{1m}^- \) are negligible (ie, \( e^+ \) and \( e^- \) are close to zero), EPT based on phase only is possible. Thus, Eq. A2 can be rewritten as (using \( \nabla^2 e^{i\phi_\pm} = e^{i\phi_\pm} \nabla i\phi_B \cdot \nabla i\phi_B + e^{i\phi_\pm} \nabla^2 i\phi_B \)):

\[
\sigma \approx \frac{1}{2\mu_0\Omega} \nabla^2 (\phi_+ + \phi_-) = \frac{1}{2\mu_0\Omega} \nabla^2 \phi_0.
\]

\[
(A3)
\]

Appendix B: Minimization to Obtain Optimal Zero-Order Phase for Multichannel Signal Combining

For breast imaging using a single transmit QBC coil, \( e^+ \) can be neglected since the spatial variations of \( B_{1m} \) are relatively small. When multiple channel receiver coils with individual coils having varying sensitivities are employed, \( e^- \) can have an influence on Eq. A2. Our goal is to minimize this \( e^- \) term by combining the multicoil signals in order to produce an effective \( B_{1m}^- \) map such that the spatial variations of \( \nabla B_{1m}^-/B_{1m}^- \) are small (ie, \( e^- \) can be neglected):

\[
\min \left\| \nabla B_{1m}^- \right\|_{B_{1m}} \quad (B1)
\]

In order to combine the individual coil signal such that a homogeneous \( B_{1m}^- \) map is produced, a complex sum with zero-order phase \( (\phi_{0,j}) \) applied to each coil \((j=1,N)\) where \( N \) is the number of receive coils) is used as a signal combining method. The combined signal \( S_{comb} \) can be represented as:

\[
S_{comb} = \sum_{j=1}^{N} S_j e^{i\phi_{0,j}} \quad (B2)
\]

where \( S_j \) is the signal at the \( j \)-th receiver coil. When including the general representation of the MR signal reception, the above equation can be rewritten as:

\[
S_{comb} = |f(B_{1m}^+)| e^{i\phi_0} \sum_{j=1}^{N} B_{1m}^+ e^{i\phi_{0,j}} \quad (B3)
\]

where \( B_{1m}^+ \) is the complex receive field at the \( j \)-th receiver coil, \( f(\cdot) \) is a \( B_{1m}^+ \) related function of the MR signal, and \( C \) is the tissue contrast term. Neglecting \( C \), we can observe the following with regard to Eqs. B1 and: (A3)

\[
\arg \min_{\phi_0=(\phi_{0,1},...\phi_{0,N})} \left\| \nabla S_{comb} \right\|_2 \approx \arg \min_{\phi_0} \left\| \nabla f(B_{1m}^+) + \nabla B_{1m}^- \right\|_2 \quad (B4)
\]

Hence, the solution to Eq. B1 can be determined by calculating the following Eq. B5. The set of optimal \( \phi_{0,j} \) was determined by:

\[
\min \left\| \nabla S_{comb} \right\|_2 \quad (B5)
\]

Since \( C \) varies in the in vivo case, a reference scan using a homogeneous phantom was employed to predetermine
the set of zero-order phase values needed to solve Eq. B5. The corresponding phase set $\phi_{0,j}$ is used for in vivo breast conductivity reconstruction.

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References