MR-Based Conductivity Imaging Using Multiple Receiver Coils

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Purpose: To propose a signal combination method for MR-based tissue conductivity mapping using a standard clinical scanner with multiple receiver coils.

Methods: The theory of the proposed method is presented with two practical approaches, a coil-specific approach and a subject-specific approach. Conductivity maps were reconstructed using the transceive phase of the combined signal. The sensitivities of the coefficients used for signal combination were analyzed and the method was compared with other signal combination methods. For validation, multiple receiver brain coils and multiple receiver breast coils were used in phantom, in vivo brain, and in vivo breast studies.

Results: The variation among the conductivity estimates was <15% as determined by the coefficient sensitivity tests. Compared with other signal combination methods, the proposed method yielded fewer artifacts in the conductivity estimates.

Conclusion: MR-based tissue conductivity mapping is feasible when using a standard clinical MR scanner with multiple receiver coils. The proposed method reduces systematic errors in phase-based conductivity mapping that can occur due to the inhomogeneous magnitude of the combined receive profile. Magn Reson Med 000:000–000, 2015. © 2015 Wiley Periodicals, Inc.

Key words: conductivity; electrical property imaging; MREPT; phase-based EPT

INTRODUCTION

MR-based electric property imaging is an emerging technique that uses B1 maps to derive information regarding tissue electrical properties at the Larmor frequency (1–6). Whereas B1 mapping conventionally refers to its magnitude mapping, MR-based electric property imaging methods require both its magnitude and phase information.

In an MR system, the phase of the measured signal contains contributions from the transmit phase (\(\Delta B_1^r\)), receive phase (\(\Delta B_r\)), and off-resonance effects. These off-resonance effects can be reduced by using spin echo acquisition or even removed completely by retrieving the phase before off-resonance effects accumulate (ie, by determining the phase at echo time [TE] = 0). Several techniques have been presented by which ultrashort TEs can be achieved (7,8). Scanners equipped with multi-transmit radiofrequency (RF) systems are capable of separating the transmit and receive phases (9). However, in the single transmit channel clinical scanners currently in use, these capabilities are limited, because these scanners typically use a quadrature birdcage transceive coil. In these coils, the transmit and receive phases are often assumed to be identical, using a symmetry argument at frequencies of up to approximately 128 MHz (3,6).

Tissue conductivity can also be measured by phase-based electric property tomography (EPT), which does not require the separation of the transmit and receive phases (2,5). The performance has been demonstrated using a separate birdcage transmit coil and birdcage receiver coil. Of note is that the accuracy of phase-based EPT relies on the spatial homogeneity of the magnitude of \(B_1^r\) and \(B_r\). For parallel transmit systems, the homogeneity of the magnitude of \(B_1^r\) and the accuracy of the phase-based EPT can be increased by \(B_1^r\) shimming (10,11). However, single transmit systems can be susceptible to \(B_1^r\) and \(B_r\) inhomogeneities.

Most clinical scanners currently use multiple receiver (multi-RX) coils to increase signal-to-noise ratio (SNR) and enable parallel imaging capabilities. In this study, we generalized the phase-based EPT technique and applied the \(B_1^r\) shimming concept to the case with multi-RX coils. For conductivity mapping, we first address the issue of combining signals in multi-RX systems from an analytic point of view. We provide a theoretical framework for our methodology and describe two practical approaches, a coil-specific approach and a subject-specific approach, for in vivo applications. Finally, we evaluate the performance of these two approaches in phantom imaging and in vivo conductivity imaging of the breast and brain.

METHODS

Theory: Phase-Based Conductivity Mapping Using Multi-RX Coils

The electric conductivity (\(\sigma\)) and permittivity (\(\varepsilon\)) of a locally homogeneous region satisfy the Helmholz equation of magnetic fields (12):

\[
\nabla^2 \mathbf{B} = i \omega \mu_0 (\sigma + i \omega \varepsilon) \mathbf{B},
\]

where \(\mathbf{B}\) is the magnetic field, \(\omega\) is the Larmor frequency, and \(\mu_0\) is the permeability of free space. Based on the Helmholz equation, for \(B_1^r\) of a transmit coil and \(B_r\) of any coil element in a multi-RX system, the conductivity and permittivity satisfy the Helmholz equation as follows:

\[
\begin{align*}
\nabla^2 B_1^r &= i \omega \mu_0 (\sigma + i \omega \varepsilon) B_1^r, \\
\nabla^2 B_{r,j} &= i \omega \mu_0 (\sigma + i \omega \varepsilon) B_{r,j}.
\end{align*}
\]
where $B_{i,j}$ is a magnetic field from the $j^{th}$ coil element from a multi-RX system. Note that since the Laplacian operation ($\nabla^2$) is a linear operation, any linear combination of $B_i$ with the complex coefficient $c_j$ from a multi-RX system satisfies the following:

$$\nabla^2 \sum_j c_j B_{i,j} = i \omega \mu_0 (\sigma + i \omega \kappa) \sum_j c_j B_{i,j}$$  \hspace{1cm} [2]$$

As suggested in MREPT (1,3), electrical conductivity can be reconstructed from the magnetic field $B_i$ of the transmit coil. This relationship can be further decomposed into the $B_i$ magnitude and phase variations of 1) the magnitude of $B_i$ if these two variables are perfectly homogenous, this approximation describes the relationship exactly. If the magnitude of the combined receive-profile $B_{comb}$ varies slowly, the conductivity can be estimated from the phase of $B_i$ in a technique referred to as phase-based EPT (4):

$$\sigma = \frac{1}{\mu_0} \mathrm{Im} \left\{ \frac{\nabla^2 B_i}{B_i} \right\}$$

$$\approx \frac{1}{\mu_0} \mathrm{Im} \left\{ \frac{\nabla^2 B_{comb}}{B_{comb}} \right\}$$  \hspace{1cm} [3]$$

A similar derivation can be made based on Equation 2. If the magnitude of the combined receive-profile ($B_{comb} = \sum_j c_j B_{i,j}$) varies slowly, the electrical conductivty can also be estimated from the phase of the combined receive profile in a multi-RX system:

$$\sigma = \frac{1}{\mu_0} \mathrm{Im} \left\{ \frac{\nabla^2 B_{comb}}{B_{comb}} \right\}$$

$$\approx \frac{1}{\mu_0} \mathrm{Im} \left\{ \frac{\nabla^2 \angle B_{comb}}{\angle B_{comb}} \right\}$$  \hspace{1cm} [4]$$

The objective is to find $c_j$ combinations that reduce the gradient of $B_{comb}$, which is the magnitude of the combined receive profile. In cases where the magnitudes of both $B_i$ and the combined receive profile vary slowly, the conductivity can be approximated as:

$$\sigma \approx \frac{\nabla^2 \{ \angle B_i + \angle B_{comb} \}}{2 \mu_0}.$$  \hspace{1cm} [5]$$

Here, the $B_i$ phase in the transmit coil and the $B_{comb}$ phase in the receive coil do not need to be explicitly separated. The accuracy of the above approximation relies on the spatial variations of 1) the magnitude of $B_i$ and 2) the magnitude of the combined receive profile resulting from the linear combination of the $B_i$ coil data. Theoretically, if these two variables are perfectly homogenous, this approximation describes the relationship exactly.

**A Practical Approach for In Vivo Conductivity Mapping**

In this study, a quadrature body transmit coil was used for RF excitation. The $B_i$ phase of the transmit coil was assumed to be slowly varying and the approximation errors due to magnitude inhomogeneity of $B_i$ in Equation 3 were not taken into account, which is consistent with phase-based EPT approaches. Thus, the goal was to minimize $\nabla B_{comb}$. It may not be possible to measure the actual $B_i$ value of a multi-RX coil in a clinical scanner. Thus we propose a practical approach for determining $c_j$ that can reduce the approximation errors due to $B_i$ variations, as given in Equation 4.

In an actual MR setting, the combined signal ($S_{comb}$) from multi-RX coils weighted by the coefficients $c_j$ can be represented by Equation 6. Furthermore, the signal from each individual coil ($S$) can be decomposed into signals that are independent of the receiver coils ($S_{common}$) and signals that are dependent on the receiver coils ($B_{i,j}$):

$$S_{comb} = \sum_j c_j S_j = S_{common} + \sum_j c_j B_{i,j} = S_{common} B_{comb}.$$  \hspace{1cm} [6]$$

To homogenize and minimize the spatial variation of the combined receive profile, $B_{comb}$, we determined the coefficients that homogenize the magnitude of $S_{comb}$ over the whole region or over a selected region of interest (ROI). Ideally, the selected ROI should not contain any spatial variation resulting from $S_{common}$ (i.e., no anatomy contrast should be present).

To select the ROI in which the required coefficients would be determined, two different approaches were used. In the first approach, a separate reference phantom examination was conducted to determine the coefficients for each multi-RX coil configuration. MR images were collected from a homogeneous phantom with no spatial contrast (constant $S_{common}$), and the coefficients were retrieved from the phantom data. The retrieved coefficients were subsequently used in vivo studies when the same receiver coil configuration was used. We anticipate that this method can be useful in breast studies, where the tissue segmentation challenges of conventional imaging methods prevents use of the subject-specific approach described below. An example would be breast imaging, where conventional imaging often has difficulty segmenting fat and parenchyma tissues. We refer to this approach as the coil-specific approach.

The second approach relied on the premise that the required coefficients could be computed directly from a homogeneous $S_{common}$ region by tissue segmentation from the acquired MR images. This approach does not require a separate phantom examination and can be used when a homogeneous ROI can be selected from the MR image. For the brain, we selected the region of gray and white matter as the ROI. This region was selected because it does not have visible contrast in the bSSFP image. For the breast, we used a region of interest that had visible contrast. The regularization parameter ($\lambda$) reflects the trade-off between the uniformity of the magnitude and the SNR of

$$\{c_j\} = \arg \min c_j \left\{ \sum_{(x,y,z) \in \text{ROI}} |S_{comb}(x,y,z) - 1|_2^2 + \lambda \sum_j |c_j|^2 \right\}.$$  \hspace{1cm} [7]$$

The regularization parameter ($\lambda$) reflects the trade-off between the uniformity of the magnitude and the SNR of...
the combined image \(S_{\text{comb}}\). This minimization, referred to as magnitude least square has been used previously to homogenize excitation angles in multitransmit systems and was solved by a variable exchange method \((13)\). Our proposed algorithm is briefly summarized as a flow diagram in Figure 1.

**Phantom and In Vivo Experiments and Reconstruction**

Phantom studies were conducted to evaluate the performance of the coil-specific approach. With in vivo breast studies in mind, four channels from one side of an eight-channel receive breast coil were used for the phantom study. Five homogeneous phantoms with different concentrations of NaCl solution \((0\%, 0.2\%, 0.5\%, 1.0\%, \text{and } 2.0\% \text{ by weight})\) were used to test the effects of variations in the reference phantom. For each phantom, the optimal coefficients were determined using Equation 7, and the multi-RX coil signals were combined. Conductivity maps were reconstructed using the phase of combined image by Equation 5, and optimal coefficients were compared. To determine the sensitivities in the conductivity maps due to differences in the coefficients, coil-specific signal combination was performed for all phantoms \((0\%, 0.2\%, 0.5\%, 1.0\%, \text{and } 2.0\% \text{ NaCl})\), and conductivity maps were reconstructed using the optimal coefficients determined from the 1.0% NaCl phantom. After reconstruction, the normalized mean square errors \((\text{NRMSEs})\) of the combined magnitude images and the average conductivity values were compared. In addition, the five sets of optimal coefficients obtained from the five test phantoms were used to reconstruct the conductivity maps for the in vivo breast data, and the conductivity values from the fat and parenchymal regions were compared.

To determine the coefficients for the in vivo bilateral breast studies, two homogeneous phantoms were built and placed in the two locations of the breast coil. The conductivity values of the right and left sides of the phantom were 2.1 S/m and 1.0 S/m, respectively, as measured by a conductivity meter \((\text{HI 8733; Hanna Instruments, Woonsocket, MA})\). These values were chosen arbitrarily; other conductivity values could also have been chosen. Using the entire phantom as the ROI, images were collected and the coefficients were determined for each side. These coefficient values were used to combine the multi-RX data sets and used for subsequent in vivo studies. In vivo experiments were performed on healthy volunteers under the approval of the local Institutional Review Board.

Phantom and breast studies for coil-specific signal combinations were performed on a GE 3T clinical scanner \((\text{MR750; GE Healthcare, Waukesha, Wisconsin, USA})\) with an eight-channel receive breast coil. The required phase for conductivity reconstruction was retrieved from multislice \(T_2^*\)-weighted fast spin echo (FSE) images that were acquired with the following parameters: number of slices \(= 50\) \((3\text{-mm thickness, no gap})\); image size \(= 416 \times 256\); field of view \(= 320 \times 320\) mm; repetition time \(= 4420\) ms; effective TE \(= 102\) ms; \(\text{(Echo Train Length)} = 20\); and total scan time \(= 2\) min.

The subject-specific approach was first tested using the phantom experimental data. For each NaCl concentration, the optimal coefficient was determined using the phantom as the segmented ROI. Next, the NRMSE of the combined image and the conductivity values were compared with the true values. Thus, this method takes an equivalent approach to the coil-specific method in that the optimal coefficients are obtained individually. In vivo brain studies were performed on a Siemens 3T Tim Trio system \((\text{Siemens Medical Solutions, Erlangen, Germany})\) with a 12-channel receive head coil. A 3D balanced steady state free precession \((\text{bSSFP})\) \((\text{TrueFISP})\) sequence was used to retrieve the required phase for conductivity reconstruction using the following parameters: spatial resolution \(= 1 \times 1 \times 2\) mm; field of view \(= 256 \times 256 \times 32\) mm; repetition time \(= 4\) ms; TE \(= 2\) ms; and total scan time \(= 4\) min, 23 s \((16\) averages). By thresholding the magnitude image of the bSSFP images, the gray and white matter region was chosen as the ROI to find the optimal coefficients.
In all cases, the conductivity was reconstructed using only the phase of the combined images, as in phase-based EPT (Eq. 4). To reduce the error resulting from boundary artifacts and the statistical noise when computing the Laplacian of the phase, the phase was first fitted to a second-order weighted polynomial model over the whole kernel (14). The Laplacian of the phase was then estimated from the second-order coefficients of the fitted polynomial. In addition, a weighting factor was determined during polynomial fitting from the magnitude of the FSE (for breast) and the bSSFP (for brain) images using Gaussian radial basis function kernel (15). Specifically, at each pixel $r_0$, the weighting factor, $w(r)$, inside the fitting kernel, $\Omega$, was determined by

$$w(r) = G_\delta(|S_\Omega(r) - S(r_0)|),$$

where $G_\delta$ is a Gaussian distribution with standard deviation $\delta$ and $S_\Omega$ refers to the magnitude of the pixel inside the fitting kernel normalized by the maximum intensity of the image. Hence, the electrical conductivity value at a given pixel was estimated from the phases of several pixels that have similar magnitude image intensities as that of the original pixel. The fitting kernel size was $15 \times 15$, $25 \times 25$, and $25 \times 25 \times 15$ for phantom, breast, and brain, respectively. When a 2D kernel was used, the second-order derivate in the third direction was ignored in computing the Laplacian of the phase. After the Laplacian calculation, an additional 2D bilateral filtering ($15 \times 15$ kernel) and a 2D median filtering ($2 \times 2$ kernel) was applied.

For comparison, conductivity maps were reconstructed using different multi-RX coil signal combination methods, including the complex sum method ($S_{\text{comb}} = \sum_{j=1}^{N} S_j$) and the zero-order phase correction and summation method ($S_{\text{comb}} = \sum_{j=1}^{N} e^{i\varphi_j} S_j$). To calibrate the zero-order phase differences between the receive channels, high SNR voxels were manually selected for all receive channels as an ROI. Next, for each channel, the average of the complex signals over the ROI was calculated (16). The phase of the average was chosen as the zero-order phase, $\varphi_j$, of the receive channel. Unless stated otherwise, all conductivity estimates for tissue are presented as means ± standard deviations in units of S/m.

FIG. 2. Coil-specific and subject-specific combinations for five phantoms with 0%, 0.2%, 0.5%, 1%, and 2% NaCl. (a) Comparison of the magnitudes of the optimal coefficients. (b) Comparison of the phases of the optimal coefficients. (c) NRMSEs of the combined magnitude images. (d) Reconstructed conductivity values and their true values.
RESULTS

The results from the phantoms with different NaCl concentrations (0%, 0.2%, 0.5%, 1.0%, and 2.0% by weight) using four channels from one side of an eight-channel receive breast coil are shown in Figure 2. The differences, both in magnitude and phase, of the optimal coefficients for each reference phantom are shown in Figure 2a and 2b. The coefficient magnitudes seemed to vary less than the coefficient phases. The NRMSEs of the combined images are given in Figure 2c. Using coefficients from individual test phantoms (as in the subject-specific case) yielded a smaller NRMSE value than when all the images were combined with the coefficients obtained with the 1.0% NaCl case (as in the coil-specific case). The overall NRMSE variations for all combinations were below 0.07. In Figure 2d, the reconstructed conductivity values from the coil-specific approach (i.e., using coefficients from the 1.0% NaCl test) and the subject-specific approach are provided. Both coil-specific and subject-specific approaches resulted in higher conductivity estimates than the conductivity values measured by the conductivity meter. However, our findings are consistent with other studies showing that phase-based EPT is subject to an overestimation bias (5). The coil-specific approach tended to overestimate these values compared with the subject-specific approach; however, the differences between these values were <15%.

The combined magnitudes, phase images, and reconstructed conductivity images in the right side of the breast using the five sets of optimal complex coefficients obtained from the five phantoms are shown in Figure 3. The inhomogeneity in the combined magnitude image near the fat region is clearly visible in the first image, suggesting an improper signal combination. However, the image quality seems comparable for the other four cases. Moreover, the reconstructed conductivity images also showed similar image quality. The reconstructed conductivity values for parenchyma were 0.32 ± 0.24 (using coefficients from the phantom calibrated with 0% NaCl), 0.40 ± 0.12 (0.2% NaCl), 0.41 ± 0.15 (0.5% NaCl), 0.45 ± 0.14 (1.0% NaCl), and 0.42 ± 0.12 (2.0% NaCl). For fat, the reconstructed conductivity values were −0.04 ± 0.27 (0% NaCl), 0.05 ± 0.17 (0.2% NaCl), 0.04 ± 0.17 (0.5% NaCl), 0.08 ± 0.11 (1.0% NaCl), and 0.12 ± 0.15 (2.0% NaCl).

The effects of $|B_1^*|$ homogeneity on phase-based EPT for the phantom study are shown in Figure 4. The combined MR images and reconstructed conductivity estimates of the two saline water phantoms acquired by a breast coil using different combination methods are also shown in Figure 4. The combined magnitude image (first row) is much more homogeneous using our proposed method, which validates the assumptions made in Equation 4. The conductivity estimates over the right water phantom (true value: 2.1) were 1.66 ± 0.42 (complex sum), 1.82 ± 0.25 (zero-order phase correction and summation), and 2.08 ± 0.19 (proposed). The conductivity estimates over the left water phantom (true value: 1.0) were 0.92 ± 0.12 (complex sum), 1.05 ± 0.18 (zero-order phase correction and summation), and 1.04 ± 0.05 (proposed). Thus our proposed method yielded the smallest standard deviation of the conductivity estimates. In addition, the artifacts in the conductivity estimates were greatly reduced, especially in the corner areas, by our proposed method. Finally, the bottom left image shows a conductivity map estimate produced by simple averaging of the conductivity maps obtained from each coil. This map illustrates the need for a signal combination algorithm.

The conductivity maps of breast tissue that were reconstructed using the coil-specific coefficients for
signal combination determined by the reference phantom study are shown in Figure 5. For our proposed method, the magnitude of the combined FSE image was more homogeneous and lacked the shading-like artifacts seen in the other methods. Furthermore, the boundaries between the breast parenchyma and fat in the conductivity map matched well with the boundaries in the FSE image. However, for the complex summation and zero-order phase correction and summation methods, the boundaries of the breast parenchyma were not clear in the conductivity maps. Moreover, the conductivity maps showed a slowly varying pattern (white arrow). The conductivity estimates over fat were 0.08 ± 0.32 (complex sum), 0.15 ± 0.25 (zero-order phase correction and summation), and 0.06 ± 0.30 (proposed). The conductivity estimates over parenchyma were 0.37 ± 0.18 (complex sum), 0.38 ± 0.12 (zero-order phase correction and summation), and 0.42 ± 0.10 (proposed). For all three methods, the averages of the estimated conductivity values were close to the conductivity values of ex vivo measurements (0.04 for fat and 0.34 for breast parenchyma) (17).

The conductivity maps of brain tissue that were reconstructed using the subject-specific coefficients for signal combination are shown in Figure 6. The conductivity map from our proposed method appears more reliable than those from the other combination methods. Over the cerebrospinal fluid (CSF) of the brain, the conductivity estimates were 1.96 ± 0.25 (complex sum), 2.20 ± 0.35 (zero-order correction and summation), and 2.16 ± 0.32 (proposed). Over the GM and WM of the brain, the conductivity estimates were 0.68 ± 0.29 (complex sum), 0.67 ± 0.49 (zero-order correction and summation), and 0.40 ± 0.13 (proposed). The conductivity values over CSF and WM using our proposed method exhibited smaller standard deviations, and the mean values were closer to the reference values (2.14 for CSF, 0.54 for GM, and 0.34 for WM) (18).

**DISCUSSION**

In this study, we developed an MR-based conductivity imaging method for standard clinical MR scanners. Our proposed method uses a multi-RX coil configuration that is found in many current scanners. While most approaches for data combination using a multi-RX coil have been developed to optimize the SNR (19), our proposed data combination method aims to reduce errors in conductivity estimates that are due to the inhomogeneous receive profile of the combined image. Our approach for reducing these errors is to determine the coefficients of combination such that the magnitude of the combined image is more homogeneous. For example, the proposed method shows a more homogeneous magnitude image over the breast parenchyma (Figure 5). On the other hand, the complex summation and zero-order phase correction and summation methods may introduce artifacts and inaccuracies in the conductivity estimates.
The reconstructed magnitude images obtained with our proposed method of data combination were more homogeneous than those constructed by other methods; moreover, the artifacts that are often evident in conductivity reconstructions were reduced. However, our proposed approach does not remove the errors that are inherent in the phase-only method (5,20).

In the coil-specific approach, a separate reference phantom study was conducted to determine the coil combination coefficients. The phantom and in vivo breast imaging results indicate the superior performance of this coil combination method compared with the other combination approaches. Tests with different calibration phantoms indicated that the variations of the optimal coil combination coefficients are within reasonable range for conductivity mapping. The phantom studies showed that the coefficients do not need to be significantly altered as long as the coil configuration does not change. A similar algorithm has been applied to differentiate between breast tissues (21).

We have proposed a subject-specific approach that does not need a preliminary phantom scan for cases in which an ROI with no contrast can be segmented (ie, the region is dominantly weighted by the receiver sensitivity). Here, as long as the image contrast is minimal, the segmented ROI may include several types of tissue. Thus, there is no need to collect MR images with good tissue contrast or to use sophisticated segmentation methods. As shown in Figure 6a, gray and white matter contrast was not visible in the TrueFISP image of the brain. Thus, the region containing the gray and white matter was segmented and used as the ROI. The coefficients can be readily obtained from this segmented ROI.

We tested this approach on a brain scan and found that this approach yielded improved performance compared with the other coil combination approaches. As shown in phantom results (Fig. 2c), this approach can achieve more uniform magnitude of combined image than the coil-specific approach. However, this approach may not achieve higher SNR in the combined image than the coil-specific approach and may result in larger errors by statistical noise, as we observed a larger standard deviation in conductivity estimates from the phantom of NaCl 2% (Fig. 2d). Although this method produced more accurate results than the coil-specific approach (Fig. 2), segmentation problems could limit its use.

For conductivity reconstruction, a weighted polynomial fitting algorithm was used before the required Laplacian was determined. During the weighting process, the contrast of the reference image determines the weighting factors and thus the effective kernel. In general, this approach produces conductivity maps with more visible boundaries compared with other filtering techniques. Hence, the maps appear to have more structural information, although the accuracy of this information cannot be guaranteed. When the conductivity variation and the reference contrast do not correlate with one another, boundary artifacts remain in the estimated conductivity. These artifacts can spread to neighboring voxels, depending on the effective kernel. Many studies have focused on how to determine the most suitable algorithm for solving the Laplacian and how to develop methods for overcoming these boundary artifacts (9,22–28). A detailed analysis is presented in the Appendix.

In our studies using a conventional body transmit coil at 3T, we assumed that the magnitude of $B_1^+$ was...
homogeneous and that the errors due to its spatial variation were negligible. Our proposed approach was not developed to mitigate errors due to inhomogeneous $|B_1^+|$ and thus might not overcome the systematic artifacts in phase-based EPT, especially for high-field MRI (e.g., 7T). Some studies have found that an additional step to shim the $B_1^+$ value can be beneficial (10,11). For 3T or less, especially in the brain, we believe that our proposed method can be applied directly.

In our proposed method, a single set of optimal coefficients is determined and then applied globally (i.e., in a spatially consistent manner). Using a single set of global coefficients might not efficiently homogenize $|B_1^+|$ for the whole breast region. Moreover, conductivity measurements could potentially have large artifacts at the voxels in the remaining regions of heterogeneity. The regions indicated by red arrows in Figure 5 denote voxels with conductivity estimates that do not match well with the literature value of breast fat (~0.04 S/m). To globally mitigate the heterogeneity of $|B_1^+|$, the ROI used for segmentation should not be localized. Also, a locally optimized signal combination method (29) could further reduce these errors when conductivity values are estimated for a local region such as a tumor region.

CONCLUSION

We have proposed a conductivity imaging method using standard clinical MR scanners with body transmit coils and multi-RX coils. Using either a coil-specific or a subject-specific signal combination method for the multi-RX coil channels, we reconstructed conductivity maps from the phase of the combined image. Moreover, we used phantom and in vivo experiments to show that our proposed signal combination method is more effective in reducing artifacts in conductivity estimates due to combined receive profiles with heterogeneous magnitudes than other simpler signal combination methods (complex summation and zero-order phase correction and summation).

APPENDIX

Boundary Artifacts in Weighted Polynomial Fitting in Phase-Based Conductivity Imaging

To demonstrate the merits and limitations of weighted polynomial fitting in phase-based conductivity imaging reconstructions, simulation studies were performed for a numerical two-compartment cylindrical phantom. Here, we did not consider multireceiver coils and signal combination from multi-RX data, which was the major focus of this study, but only concentrated on the case of reconstructing conductivity from the phase of one $B_1^+$ map to evaluate the boundary artifacts using weighted polynomial fitting methods.

Electromagnetic simulations were performed using the Bessel boundary matching method (30). A two-compartment numerical phantom with radii of 0.1 and 0.2 m was designed to have infinitely long cylinders. The conductivity values of the compartments were 0.5 and 1 S/m, as shown in Figure A1a. The location of the

![Fig. A1. Conductivity estimates of two compartment numerical model from simulated complex $B_1^+$ map with three levels of SNR of 10,000, 1000, and 200. (a) Original conductivity of the model. (b) Conductivity estimates using five-point discrete Laplacian. (c) Conductivity estimates using 2D polynomial fitting. (d) Conductivity estimates using weighted 2D polynomial fitting. The weighting factors were determined from the magnitude image with contrast of 2:1, as shown in Figure A2a. Note that no postfiltering scheme was applied here (b–d), contrary to the experimental data in Figures 2–6.](image_url)
RF coil and shields were fixed at radii of 0.3 and 0.35 m and were driven by an eight-rod quadrature coil at 128 MHz (3T). The axial slice of the $B_1^+$ map was simulated with a spatial resolution of $1 \times 1$ mm. Complex Gaussian noise was added to the $B_1^+$ map before phase-based conductivity reconstructions.

Three types of reconstruction techniques were compared: five-point discrete Laplacian operator, 2D polynomial fitting, and our proposed 2D weighted polynomial fitting using a Gaussian radial basis function kernel (15). For both polynomial fitting methods, a $15 \times 15$ kernel was applied, and the conductivity value was estimated from the second-order coefficients of the fitted polynomial. For weighted polynomial fitting, the weighting factors were determined by Equation 8 using magnitude images. The same standard deviation of $\delta = 0.05$ in Gaussian kernel was used as in the breast conductivity imaging. Four different contrasts (2:1, 1.4:1, 1.2:1, and 1:1) in the magnitude images were used to evaluate the performance in terms of boundary artifacts.

As shown in Figure A1b–A1d), polynomial fitting-based reconstructions were much more robust to the statistical noise than five-point discrete Laplacian. For polynomial fitting, the boundary artifacts were broadened according to the fitting kernel size. However, using weighted polynomial fitting, the boundary artifacts were reduced greatly. Here, the weighting factors were computed from the magnitude image with 2:1 contrast as shown in Figure A2a.

Figure A2 shows the sensitivity of the conductivity maps with respect to the contrast in the magnitude images, which is used to determine the weighting factors. As the contrast increases, the magnitude and the width of boundary artifacts are reduced. With no contrast (ie, contrast = 1:1), the conductivity estimate matched the reconstructed conductivity exactly using the polynomial fitting shown in Figure A1b.

**REFERENCES**