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Determining Electrical Properties Based on B_1 Fields Measured in an MR Scanner Using a Multi-channel Transmit/Receive Coil: a General Approach

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Abstract

Electrical Property Tomography (EPT) is a recently developed noninvasive technology to image the electrical conductivity and permittivity of biological tissues at Larmor frequency in Magnetic Resonance (MR) scanners. The absolute phase of the complex radio-frequency (RF) magnetic field (B_1) is necessary for electrical property calculation. However, due to the lack of practical methods to directly measure the absolute B_1 phases, current EPT techniques have been achieved with B_1 phase estimation based on certain assumptions on object anatomy, coil structure and/or electromagnetic wave behavior associated with the main magnetic field, limiting EPT from a larger variety of applications. In this study, using a multi-channel transmit/receive coil, the framework of a new general approach for EPT has been introduced, which is independent on the assumptions utilized in previous studies. Using a human head model with realistic geometry, a series of computer simulations at 7T were conducted to evaluate the proposed method under different noise levels. Results showed that the proposed method can be used to reconstruct the conductivity and permittivity images with noticeable accuracy and stability. The feasibility of this approach was further evaluated in a phantom experiment at 7T.

1. Introduction

Noninvasive imaging of tissue electrical properties, including conductivity and permittivity, has drawn great interest among researchers in the past decades. In the frequency range of ~1 kHz to ~10² MHz, electrical properties have been demonstrated in several studies to be significantly altered as a function of tissue malignancy (Joines 1994, O'Rourke *et al* 2007, Swarup *et al* 1991), holding promise for detection and characterization of tumors (Fear *et al* 2002). Efforts have been made to develop imaging methodologies for noninvasive characterization and imaging of electrical properties, including Electrical Impedance Tomography (EIT) (Paulson *et al* 1993, Metherall *et al* 1996), Magnetic Resonance Electrical Impedance Tomography (MREIT) (Zhang 1992, Kwon *et al* 2002, Gao *et al* 2005, Lee *et al* 2011) and Magnetoacoustic Tomography with Magnetic Induction (MAT-MI) (Xu and He 2005, Li *et al* 2007, Xia *et al* 2007, 2010, Hu *et al* 2010, Mariappan *et al* 2011). More recently, a new technology, named Electrical Properties Tomography (EPT) (Haacke *et al* 1991, Wen 2003, Katscher *et al* 2009, Zhang *et al* 2010), has been introduced to

noninvasively image tissue conductivity and permittivity at proton Larmor frequency, based on complex-valued maps of radiofrequency (RF) field (B_1) measured in MRI experiments. Compared with other relevant technologies, EPT offers the advantages of especially fairly high spatial resolution, using non-ionizing radiation, and simplified operation on subjects.

From another perspective, RF heating of tissue is a major safety concern in MR imaging at high field (3T) (Gandhi and Chen 1999), even more so at ultra-high field (7T) MRI (Collins *et al* 2004), and guidelines have been established to not exceed pre-defined limits for global and local Specific Absorption Rate (SAR). Computing local SAR however requires precise knowledge of local electric fields, tissue conductivity and mass density. Until now, this information has been obtained through model-based and highly time-consuming numerical simulations; furthermore, such electromagnetic models have been developed for only a small number, i.e. in the order of a few tens, of different human anatomies (Voigt *et al* 2011).

An alternative option offered by EPT is to map *in vivo*, for a given subject, electrical tissue properties and electrical field distributions based on MR acquisitions, holding promise to effectively calculate local SAR on a subject-specific basis for any RF pulse design. When successfully established, the EPT approach could lead to major advancement towards achieving real-time subject-specific SAR management.

EPT requires a reliable measurement of complex B_1 field maps (magnitude and phase) to calculate electrical tissue properties. Although the absolute magnitude of transmit B_1 fields (B_1^+) can be accurately measured on an MR scanner, the signal phase measured with coil elements, that are used for both transmission and reception ("transceivers"), is typically a combination of transmit, receive and system phase components (Van de Moortele *et al* 2005). Still, *relative* B_1^+ and *relative* receive B_1 (B_1^-) phases between channels can be accurately measured, taking any channel as a reference, but *absolute* B_1^+ and B_1^- phases are not accessible via a straightforward measurement. In most previous EPT studies, using a quadrature coil, B_1^+ phase was simply estimated as roughly one half of the measured transceiver phase, with the transceiver phase representing the sum of B_1^+ phase and B_1^- phase (Wen 2003, Katscher *et al* 2009, Voigt *et al* 2011, van Lier *et al* 2012). However, the underlying assumption that B_1^+ phase and B_1^- phase are approximately equal quantities cannot be sustained as the field strength increases and/or when a non-quadrature coil is used (Van de Moortele *et al* 2005).

Recently, the concept of reconstructing transmit/receive phase using multiple independent transmit/receive channels has been introduced (Zhang *et al* 2012, Katscher *et al* 2012). As shown in figure 1, for EPT using a multi-channel transmit/receive coil, the sample receives RF excitation through the transmit B_1 field ($\tilde{B}_{1j}^+$) of channel j , and the emitted MR signal is picked up through the conjugate of receive B_1 field ($\tilde{B}_{1k}^{-*}$) of channel k . Note that the magnitude of the measured signal is also proportional to the sample proton density ρ ; as a result, the *proton density weighted* receive field $\rho\tilde{B}_{1k}^{-*}$, rather than $\tilde{B}_{1k}^{-*}$ is typically mapped in MR experiments. In the notations above, ' $\sim$ ' denotes a complex quantity and '*' the complex conjugate. Zhang *et al* (2012) reported EPT using a multi-channel transceiver coil at 7T, taking advantage of mirroring symmetry about the YZ-plane (see figure 1) observed between transmit and receive B_1 field for an electrically symmetric sample and coil (Van de Moortele *et al* 2005, Van de Moortele and Ugurbil 2009). Katscher *et al* (2012) also proposed utilizing multiple transmit channels to reconstruct absolute B_1 phase for SAR estimation at 3T, using a global polynomial approximation of B_1 phase distribution.

In the current study, we introduce the framework of a new general approach to calculate conductivity and relative permittivity ϵ_r , only utilizing quantities that are measurable in an

MR scanner, i.e. magnitude and relative phase of B_1^+ and B_1^- from multiple channels. Assuming the electrical properties of sample to be piecewise homogeneous, this approach does not depend on assumptions such as sample symmetry, phase distribution, coil structure or field strength. Simulation studies were conducted at 298 MHz (the Larmor frequency of protons at 7T) to evaluate this approach under different noise levels, and experimental feasibility was demonstrated in a phantom experiment at 7T.

2. Theory

Ignoring the relatively small deviation of magnetic permeability of biological tissue from that of the vacuum, the time-harmonic Helmholtz equation for $\tilde{B}_{1j}^+$ within electrically piecewise homogeneous region can be written as (Wen 2003, Voigt *et al* 2011, Zhang *et al* 2012, van Lier *et al* 2012)

$$-\frac{\nabla^2 \tilde{B}_{1j}^+(\mathbf{r})}{\tilde{B}_{1j}^+(\mathbf{r})\omega^2\mu_0} = \varepsilon_r(\mathbf{r})\varepsilon_0 - i\sigma(\mathbf{r})/\omega \quad (1)$$

where i is the imaginary unit, ω the Larmor angular frequency, $\mathbf{r}$ denotes position, and μ_0 and ε_0 the permeability and permittivity of vacuum, respectively. For simplicity, the spatial coordinate variable $\mathbf{r}$ will not be mentioned through the rest of the paper. This equation is derived through the superposition of the Helmholtz equations for the Cartesian x- and y-components ($\tilde{B}_x$ and $\tilde{B}_y$) of a time-harmonic magnetic field (Wen 2003) according to the principle of reciprocity (Hoult 2000), which states that transmit $\tilde{B}_1^+$ and receive $\tilde{B}_1^-$ fields are linear combinations of $\tilde{B}_x$ and $\tilde{B}_y$. By expanding the Laplacian on the left hand side of (1) and separating real and imaginary components, the expressions of σ and ε_r become (Voigt *et al* 2011, Zhang *et al* 2012)

$$\begin{cases} \sigma = \frac{1}{\omega\mu_0} \left[\nabla^2 \phi_j^+ + 2 \frac{(\nabla|\tilde{B}_{1j}^+|)}{|\tilde{B}_{1j}^+|} \cdot \nabla \phi_j^+ \right] \\ \varepsilon_r = \frac{1}{\omega^2\mu_0\varepsilon_0} \left[-\frac{\nabla^2|\tilde{B}_{1j}^+|}{|\tilde{B}_{1j}^+|} + (\nabla\phi_j^+)^T \cdot \nabla\phi_j^+ \right] \end{cases} \quad (2)$$

We arbitrarily select channel n as the reference channel. By subtracting both hand sides of (2) between channel j and channel n , σ and ε_r vanish, and we obtain the following equation:

$$\begin{cases} \left(\frac{(\nabla|\tilde{B}_{1j}^+|)}{|\tilde{B}_{1j}^+|} - \frac{(\nabla|\tilde{B}_{1n}^+|)}{|\tilde{B}_{1n}^+|} \right) \cdot \nabla\phi_n^+ = -\frac{1}{2}\nabla^2\phi_{rj}^+ - \frac{(\nabla|\tilde{B}_{1j}^+|)^T}{|\tilde{B}_{1j}^+|} \cdot \nabla\phi_{rj}^+ \\ (\nabla\phi_{rj}^+)^T \cdot \nabla\phi_n^+ = \frac{1}{2} \left[\frac{\nabla^2|\tilde{B}_{1j}^+|}{|\tilde{B}_{1j}^+|} - \frac{\nabla^2|\tilde{B}_{1n}^+|}{|\tilde{B}_{1n}^+|} - (\nabla\phi_{rj}^+)^T \cdot \nabla\phi_{rj}^+ \right] \end{cases} \quad (3)$$

where ϕ_{rj}^+ is the *relative* transmit phase of channel j ($j \neq n$) with channel n as the reference channel. More details on how equation (3) was derived can be found in the Appendix. In (3), there are three unknowns of $\nabla\phi_n^+ : \partial\phi_n^+/\partial x, \partial\phi_n^+/\partial y$ and $\partial\phi_n^+/\partial z$. Equation (3) is linear and can be solved by measuring the magnitude and relative phase of $\tilde{B}_{1kj}^+$ from as few as three channels, of which one is taken as the reference. Careful channel selection should consider covering the considered region of interest (ROI) with sufficient signal-to-noise ratio (SNR) for B_1 measurement. One particular strategy is to choose adequate number of channels covering a specific ROI with stronger B_1 magnitude as can be observed in the measured B_1 maps of all channels. Naturally, using more channels will improve numerical accuracy and stability of the solution, as long as they contribute with sufficient SNR. Once $\nabla\phi_n^+$ is derived

from this set of equations (using three or more channels), ρ and ϕ can be calculated using (2).

In principle, a similar equation, based on the same Helmholtz magnetic equation but in the form of $\tilde{B}_{1k}^-$, could also be derived. However, only $|\rho\tilde{B}_{1k}^-|$, rather than $\tilde{B}_{1k}^-$, can be measured in practice. Thus, within restricted regions where ρ is assumed to be approximately homogeneous, equation (2) can be re-written as

$$\begin{cases} \sigma = \frac{1}{\omega\mu_0} \nabla^2 \phi_k^- + 2 \frac{(\nabla|\rho\tilde{B}_{1k}^-|)^T}{|\rho\tilde{B}_{1k}^-|} \cdot \nabla \phi_k^- \\ \epsilon_r = \frac{1}{\omega^2\mu_0\epsilon_0} \left[-\frac{\nabla^2|\rho\tilde{B}_{1k}^-|}{|\rho\tilde{B}_{1k}^-|} + (\nabla\phi_k^-)^T \cdot \nabla\phi_k^- \right] \end{cases} \quad (4)$$

Within these limits, (2) in the form of $\tilde{B}_{1j}^+$ and its equivalent (4) in the form of $\rho\tilde{B}_{1k}^-$ can even be summed to calculate ρ and ϕ utilizing the measurable transceiver phase (Van de Moortele *et al* 2005, Voigt *et al* 2011, Katscher *et al* 2012) $\phi_{jk} = \phi_j^+ + \phi_k^-$ in order to reduce artifacts that may result from cumulative numerical errors in phase calculation.

Note that compared to another algorithm (Zhang *et al* 2012), in which the z- component of the magnetic field is assumed to be negligible in order to calculate the gradient of the absolute B_1 phase, the present approach solves the phase gradients only using measurable transverse magnetic field components. Each set of selected channels determines the phase gradients voxel by voxel, without the assumption of a global phase approximation (Katscher *et al* 2012). In addition, without assumption about RF coils, it enables optimized coil constructions for an improved EPT solution in the ROI.

3. Materials and methods

3.1. Computer simulation

As illustrated in figure 2, a simulation was performed with the numerical model of a 16-channel transmit/receive array coil (Adriany *et al* 2008) loaded with an isotropic human head, which includes gray matter (GM), white matter (WM), cerebrospinal fluid (CSF), etc. The tissue electrical properties were assigned based on values reported in literature (Gabriel *et al* 1996b). Based on the finite-difference-time-domain-method (FDTD), sixteen complex transmit and receive B_1 fields, with a resolution of $2 \times 2 \times 2.5 \text{ mm}^3$, were calculated by exciting one channel at a time at Larmor frequency for protons at 7T (298 MHz), using the XFDTD software (version 6.3, Remcom Inc., PA, USA).

Four channels of complex $\tilde{B}_{1j}^+$ at a time, including one assigned as the reference channel, were used to reconstruct one set of ρ_n^+/x , ρ_n^+/y and ρ_n^+/z for the aforementioned reference channel. These quantities were inserted into equation (2) to calculate one set of parametric ρ map. In order to reduce artifacts inherently amplified by the Laplacian operator over the reconstructed phase, equations (2) and (4) were summed to obtain ρ , utilizing the additional measurable quantities $|\rho\tilde{B}_{1k}^-|$ and transceiver phase $\phi_n = \phi_n^+ + \phi_n^-$ for channel n . In order to minimize coupling between channels, every fourth channel was picked, following azimuthal geometric order, to form each set of four channels. This process was repeated with each of the sixteen channels used as the reference, yielding a total of sixteen sets of ρ and ϕ for the same head model to be combined to produce the final parametric maps. At the boundaries separating tissues with different electrical properties, artifactual errors would be induced as a direct consequence from the local violation of the piecewise homogeneous assumption. These errors, which typically yielded locally out-of-range electrical property

values, were identified by applying predefined upper and lower thresholds of valid electrical properties, and the corresponding values were recalculated by interpolation between nearby voxels.

The aforementioned reconstruction was performed after adding different levels of Gaussian white noise to the real and imaginary parts of the complex B_1 fields, with standard deviation set to 1% and 2% of the average B_1 magnitude for each channel in the ROI. The noise level was determined based on noise analysis in the experimental data. The noise-contaminated B_1 data were firstly spatially smoothed using a Gaussian low-pass filter before the Laplacian operator was applied.

To evaluate the performance of the method, we used the averaged relative error (RE) and correlation coefficient (CC) to measure accuracy and similarity between the distributions of reconstructed and target maps, respectively. For a given parameter q , RE and CC are defined as

$$RE_q = \frac{\sum_{i=1}^N \left| \frac{q_{ri} - q_i}{q_i} \right|}{N} \quad (5)$$

$$CC_q = \frac{\sum_{i=1}^N (q_{ri} - \bar{q}_r) \cdot (q_i - \bar{q})}{\sqrt{\sum_{i=1}^N (q_{ri} - \bar{q}_r)^2 \cdot \sum_{i=1}^N (q_i - \bar{q})^2}} \quad (6)$$

3.2. Experiment

Experimental data were acquired in a phantom to further evaluate the proposed algorithm. The experiments were carried out on a 7T MR system (Siemens, Erlangen, Germany; Magnex Scientific, Oxford, UK). A 16-channel transmit/receive head coil, which had been modeled for the simulation data, was utilized to transmit and receive RF energy, with the transmit channels powered by 16×1kW amplifiers (CPC, Hauppauge, NY, USA) interfaced with a remotely controlled phase/amplitude gain unit. A single-compartment cylindrical phantom was built with a diameter of 8.7 cm and a length of 20 cm. It was filled with a gel of saline solution composed of distilled water, NaCl, Gelatin (G2500, SIGMA) and $\text{CuSO}_4 \cdot 5\text{H}_2\text{O}$ in the mass ratio of 100:0.12:3:0.025. Its conductivity and relative permittivity were measured as $\sigma = 0.34 \text{ S/m}$ and $\epsilon_r = 77$ at 298 MHz, using an Agilent 85070D dielectric probe kit and an Agilent E4991A network analyzer at Pennsylvania State University College of Medicine. During the experiment, the phantom was placed in the center of the coil with its longitudinal direction parallel to B_0 field. In general, mapping transmit B_1 field has the inherent advantage of avoiding the issue of proton density bias that typically occurs when mapping receive B_1 field, where the relative contributions of proton density and B_1 magnitude $|\tilde{B}_{1k}^-|$ cannot be distinguished in the measured product $|\rho \tilde{B}_{1k}^-|$. However, in the particular case of a homogeneous phantom, as in the present study, ρ is by definition constant throughout the entire phantom, and thus, is canceled out in equation (4), making it possible to utilize measured $|\rho \tilde{B}_{1k}^-|$ and $e^{i\phi_{rk}^-}$ for reconstruction. A straightforward advantage of this approach is that a higher SNR can typically be achieved per unit of acquisition time when mapping proton density weighted receive B_1 field data. Thus, EP reconstruction was performed based on the latter, rather than on transmit B_1 information. Details of imaging protocols and sequences utilized to acquire magnitude and relative phase

maps, including $|\rho\tilde{B}_{1k}^-|$ and ϕ_{rk}^- for multiple channels can be found in a previous study (Zhang *et al* 2012). Briefly, a 3D flip angle (FA) map was measured using the Actual Flip Angle technique (AFI) (Yarnykh 2007), all channels transmitting together with RF power adjusted to reach a high flip angle regime. Subsequently, with all channels still transmitting together, GRE images (gradient-recalled echo) were obtained, using the same FA as for AFI, with long T_R and short T_E to minimize the impact of T_1 and T_2^* . The corresponding signal magnitude of the 16 receive channels was normalized by the sine of the acquired 3D FA map to generate $|\rho\tilde{B}_{1k}^-|$ for the 16 channels (β being constant) (Van de Moortele *et al* 2005). Corresponding relative phase maps were obtained from the phase of the same high SNR data set. Relevant acquisition parameters for 3D AFI: $T_E/T_{R1}/T_{R2}=3.23/20/120$ ms, nominal FA=60°, eight averages, acquisition time ~170 min, and high SNR GRE images: $T_E/T_R=3.73/5000$ ms, nominal FA=60°, 20 averages, acquisition time ~210 min. Twelve continuous slices were measured with a spatial resolution of $1.5 \times 1.5 \times 1.5$ mm³.

The acquired data were firstly smoothed with a Gaussian filter to reduce high frequency noise. During the reconstruction, four channels of $\rho\tilde{B}_{1k}^-e^{i\phi_{rk}^-}$ were used at a time to calculate $\nabla\phi_k^-$, with one of the four channels taken as a reference. In the absence of available transceiver phase, both ρ and ϕ_r were calculated using (4) once $\nabla\phi_k^-$ was obtained.

4. Results

Using the numerical simulation of the coil array loaded with a head model, we evaluated the performance of our proposed method to image the electrical property maps of soft brain tissues, including gray matter (GM), white matter (WM) and cerebrospinal fluid (CSF).

Figure 3 illustrates, within a slice of interest, reconstructed phase maps of the transmit B_1 field in comparison with the target phase maps, i.e. as obtained in the simulation data, under noise free condition. Here, the magnitude and relative phase of transmit B_1 field from channel #3, #7, #11 and #15 were sent into (3), with channel #3 as the reference. The reconstructed phase maps agreed well with simulated ones, with a mean error of 0.13 radians and a CC of 97.2%.

Under the same noise free condition, the reconstructed electrical property maps of the tissues of interest (GM, WM and CSF) are shown in figure 4 for the same slice as in figure 3, with other tissues masked by the target electrical property values around the periphery. The predefined [min, max] thresholds for removing boundary artifacts were set to [0, 3] S/m for ρ and [0, 100] for ϕ_r , based on reported values for brain tissues (Gabriel *et al* 1996a). Due to the violation of piecewise homogeneous assumption, noticeable residual error can be observed near the boundaries of tissues. Overall, the electrical properties of the brain tissues are successfully reconstructed, and in comparison with the target images, anatomical structures of GM, WM and CSF are clearly identified in the estimated maps for both ρ and ϕ_r . Within the ROIs of GM and WM, the reconstructed mean $\pm$ SD (standard deviation) of ρ are 0.84 ± 0.19 and 0.55 ± 0.08 S/m, respectively, with respect to the target values of 0.83 and 0.49 S/m; the reconstructions of ϕ_r are 54.1 ± 11.2 and 48.1 ± 10.1 , respectively, comparing to target values of 58.3 and 42.8. The imaging results under noise levels of 1% (SNR=100) and 2% (SNR=50) are shown in figure 5. A Gaussian filter with kernel size of 7 was applied to the B_1 field. As a result of the spatial filtering on the B_1 field, the reconstructed electrical property maps were spatially smoothed as we can see in figure 5(c)–(f). Thus, CSF, which covers a minor area in the original target images (figure 5(a) and (b)) is not seen in the reconstructed images. A smaller kernel size would help better define the image structure, but may compromise the accuracy due to intrinsic sensitivity of the Laplacian operator in EPT algorithms. When SNR=100, the reconstructed electrical properties in ROIs of GM and WM

are -0.81 ± 0.19 and 0.66 ± 0.14 S/m, respectively, and $\sigma_r = 57.1 \pm 7.6$ and 48.7 ± 5.8 , respectively; when SNR=50, the corresponding results are -0.81 ± 0.18 and 0.65 ± 0.14 S/m, and $\sigma_r = 57.7 \pm 7.7$ and 49.0 ± 6.5 .

In order to verify the appropriateness of using every fourth channel to generate each group of four channels to calculate EP maps, we investigated the impact of the distance between channels within a group by comparing RE in calculated σ and σ_r maps when using every first, second, third or fourth channels to form each group. As shown in figure 6, keeping the maximum distance of 4 coil elements between channels within a group clearly provided the smallest residual errors. This can be explained by the fact that the electromagnetic coupling between coil elements, which are utilized in the computation, decreases as the distance between them increases.

For the experimental study, the proton density weighted receive B_1 field $\rho \tilde{B}_{1k}^- e^{i\phi_{rk}^-}$ was utilized to obtain the solution based on (4). To demonstrate the validity of the proposed method, we first evaluated B_1 phase reconstruction on experiment data. Due to lack of knowledge about the absolute B_1 phase information, we compared the reconstructed relative phase of receive B_1 field to the measured relative phase in a transverse slice between two arbitrary channels. Shown in figure 7(a) is the measured relative phase of receive B_1 field between channel #4 and #15 in slice 6 while figure 7(b) shows the calculated relative phase between the same two channels, which was calculated as the difference of the reconstructed absolute phase maps of these two channels using two independent groups of four channels (group 1: channels #4 (reference), #8, #12 and #16; group 2: channels #3, #7, #11 and #15 (reference)). Strong consistency is observed between the measurement and the reconstruction, with CC = 97.6% and an average difference of 0.08 radians.

Line profiles through the reconstructed σ and σ_r maps along the x-direction, crossing the center of the phantom, are shown in figure 8. In the region where FA was greater than 36° (providing high SNR), which covers more than half of the slice and includes the entire central profile, the reconstruction results (mean $\pm$ SD) are -0.33 ± 0.08 S/m and $\sigma_r = 78.2 \pm 5.4$. As a reference, the aforementioned values of σ and σ_r measured by a probe are 0.34 S/m and 77, respectively. Meanwhile, the reconstruction results throughout the entire slice are -0.39 ± 0.14 S/m and $\sigma_r = 71.0 \pm 14.0$.

5. Discussion and Conclusion

In the present study, a general approach for calculating conductivity and permittivity, using the B_1 field data from multiple transmit/receive channels measured in an MR scanner, was introduced. Using simulation data based on the model of a multi-channel transmit/receive coil loaded with a human head at 7T, feasibility has been shown for imaging electrical properties of human tissue under both noise-free and noisy situations. Furthermore, the validity of the approach was experimentally demonstrated in a phantom on a 7T MR scanner.

Until now, lack of measurable absolute phase for transmit/receive B_1 field has imposed serious limitations on EPT approaches. Thus, various assumptions for coil, sample and field strength have been necessary to meet certain condition for the B_1 field in order to solve the Helmholtz equation (Voigt *et al* 2011, Zhang *et al* 2012, van Lier *et al* 2012). In the present study, the framework of the proposed method does not rely on any of these assumptions, offering the potential of significantly broadening the range of applications for EPT. For example, with flexible channel selection and unconstrained field distribution, EPT could contribute as a diagnostic tool for characterizing malignant liver tissue (O'Rourke *et al* 2007), while other EPT approaches, which depend on symmetry of the sample or equal-

phase assumption, may suffer from asymmetry or large scale of the abdomen. Without limitation on field strength, EPT can potentially benefit from higher magnetic fields taking advantage from the more rapid spatial change of B_1 fields and from improved SNR of the acquired B_1 data. In turn, improved EPT could provide reliable real-time and patient-specific SAR calculation, which is especially significant at ultra-high field.

Different from quadrature coils, where transmit and receive B_1 fields exhibit fixed spatial patterns with very limited degrees of freedom, RF coils with multiple transmit/receive channels can shape the B_1 field distribution dynamically for imaging benefit; in addition, multi-channel B_1 magnitude and relative phase between channels can be obtained, after which the non-measurable absolute phase of each channel can be derived through different ways. The method proposed by Zhang *et al* (2012) was based on the observation of mirroring relationship between transmit and receive B_1 magnitude and on neglecting the B_z component to obtain the phase solution, requiring both within-plane object symmetry and channels aligned in parallel with the z-direction. In the study by Katscher *et al* (2012), the absolute phase was estimated based on a global polynomial approximation of the phase distribution throughout the region of interest. However, errors that resulted from eliminated higher order components of the spatial phase distribution can propagate into the calculation of electrical properties and SAR through subsequent spatial derivation. By contrast, the method proposed here can calculate the absolute phase without dependency on sample symmetry, coil structure or distribution of the B_z component.

With the formalism of the proposed approach, it is the first time that only the magnitude and the relative phase of transmit or receive B_1 fields are sufficient to calculate the parametric maps of the desired electrical properties and accurately estimate the corresponding absolute B_1 phase. This approach merits further investigation in the context of real time, subject-specific SAR estimation during an MR scanning session. However, in the simulation study performed in the current work, in order to avoid Laplacian operation on the calculated absolute transmit or receive B_1 phase, the transceiver phase was used to calculate the conductivity. In actual experiments, this will lead to additional scanning time because the transceiver phase and information on both transmit and receive B_1 fields are needed when equation (2) and (4) are summed to calculate the conductivity. Further optimization in acquisition methods, however, can help addressing this issue. For example, the transceiver phase could be measured using faster imaging sequences while providing satisfactory SNR (Stehning *et al* 2011). On the other hand, investigation on imaging processing tools on the retrieved B_1 phase will be useful to eliminate using the transceiver phase.

The boundary effect due to violation of homogeneous assumptions is the major source of residual errors in the proposed method. Although boundary errors were effectively suppressed by using predefined thresholds on the parametric maps, as shown in the simulation study, valuable information may be lost under exceptional situation such as a tumor with electrical properties significantly differing from that of surrounding tissues. More advanced filtering algorithms could be developed to address boundary errors arising in calculated phase gradient maps. In addition, the mathematical formalism allowing for the inhomogeneous version of the Helmholtz equation (Zhang *et al* 2010) could be further developed in order to be integrated within the general theoretical framework proposed here. Meanwhile, it is worth noting that in the simulation study, a piecewise homogeneous head model was used with discrete discontinuities of electrical properties at tissue boundaries; it remains to be seen whether *in vivo* experimental data may provide smoother transition of electrical properties, that would inherently mitigate the aforementioned boundary issues.

Directly benefiting from high spatial resolution accessible with MRI data, in completely non-invasive settings, MR-based EPT methods hold strong promises towards *in-vivo*

characterization of tissue electrical properties. In this study, a new general framework was introduced to further enable the development of EPT methods with the ultimate goal of enabling clinical EPT applications.

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Appendix. Derivation of equation (3) from equation (1) in an electrically piecewise-homogeneous medium

Expanding the Laplacian operation on the complex B_1 field $\tilde{B}_{1j}^+ = |\tilde{B}_{1j}^+| e^{i\phi_j^+}$ of channel j , we have

$$\nabla^2 \tilde{B}_{1j}^+ = e^{i\phi_j^+} \left[\nabla^2 |\tilde{B}_{1j}^+| + 2i(\nabla |\tilde{B}_{1j}^+|)^T \cdot \nabla \phi_j^+ + i|\tilde{B}_{1j}^+| \nabla^2 \phi_j^+ - (\nabla \phi_j^+)^T \cdot (\nabla \phi_j^+) |\tilde{B}_{1j}^+| \right] \quad (\text{A.1})$$

Now dividing equation (A.1) by $-\tilde{B}_{1j}^+ \omega^2 \mu_0$, equation (1) can be rewritten as

$$\begin{aligned} \epsilon_r \epsilon_0 - i\sigma/\omega \\ = \frac{1}{\omega^2 \mu_0} \left[-\frac{\nabla^2 |\tilde{B}_{1j}^+|}{|\tilde{B}_{1j}^+|} + (\nabla \phi_j^+)^T \cdot (\nabla \phi_j^+) \right] - i \frac{1}{\omega^2 \mu_0} \left[\nabla^2 \phi_j^+ + 2 \frac{(\nabla |\tilde{B}_{1j}^+|)^T}{|\tilde{B}_{1j}^+|} \cdot \nabla \phi_j^+ \right] \end{aligned} \quad (\text{A.2})$$

Separating the real and imagery components in equation (A.2), equation (2) can be derived immediately.

Given a reference transmit channel n , the absolute transmit B_1 phase of any transmit channel j can be expressed as:

$$\phi_j^+ = \phi_{rj}^+ + \phi_n^+ \quad (\text{A.3})$$

Based on (A.3), we can further expand the phase-related items in equation (2) and get

$$\begin{aligned} \sigma = \frac{1}{\omega \mu_0} \left[\nabla^2 \phi_{rj}^+ + \nabla^2 \phi_n^+ + 2 \frac{(\nabla |\tilde{B}_{1j}^+|)^T}{(|\tilde{B}_{1j}^+|)} \cdot (\nabla \phi_{rj}^+ + \nabla \phi_n^+) \right] \\ \epsilon_r = \frac{1}{\omega^2 \mu_0 \epsilon_0} \left[-\frac{\nabla^2 |\tilde{B}_{1j}^+|}{|\tilde{B}_{1j}^+|} + (\nabla \phi_{rj}^+)^T \cdot \nabla \phi_{rj}^+ + 2(\nabla \phi_{rj}^+)^T \cdot \nabla \phi_n^+ + (\nabla \phi_n^+)^T \cdot \nabla \phi_n^+ \right] \end{aligned} \quad (\text{A.4})$$

At a specific position, σ and ϵ_r do not depend on channels. Subtracting equation (A.4) for channel n from equation (A.4) for channel j , we obtain equation (3).

Similar derivation can be applied to acquire the corresponding equations based on receive B_1 field.

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Figure 1.

Schematic diagram of EPT. The sample to be mapped (bean-like shape in the figure) is positioned in the scanner inside a N-channel transceiver RF coil, with each coil element aligned on a circle (dotted line) in this example. For clarity, one channel j ($j \in [1, \dots, N]$) is shown in transmit mode while one channel k ($k \in [1, \dots, N]$) is shown in receive mode. The long axis of the coil elements is parallel to the main magnetic field (Z axis) and perpendicular to the figure plane. During transmission, each channel (e.g. j) excite the sample with transmit field $\tilde{B}_{1j}^+(\mathbf{r})$, whereas during reception the signal measured on each channel (e.g. k) is proportional to the product of proton density (ρ) by receive field $\tilde{B}_{1k}^-(\mathbf{r})$.

Figure 2.
 (a) 3D view of the head model and coil elements. (b) Transverse view of the simulation model and numbering order of the coil element.

(a)

(b)

Figure 3.
 (a) Simulated phase maps of transmit B_1 field for 16 channels. (b) Reconstructed phase maps of transmit B_1 field for corresponding channels.

Figure 4. Reconstruction of σ and ϵ_r using the proposed algorithm applied on simulation data at 7T under noise free condition. (a)–(b) Target electrical properties. (c)–(d) Reconstructed properties.

Figure 5.

Reconstruction of σ and ϵ_r using the proposed algorithm applied on simulation data at 7T under different noise levels. (a)–(b) Target electrical properties. (c)–(d) Reconstructed properties with SNR=100. (e)–(f) Reconstructed properties with SNR=50.

Figure 6.

Mean REs for reconstructed EP maps as a function of distance between neighboring channels expressed as number of coil elements. (a) Conductivity. (b) Relative permittivity.

Figure 7.

Experimental evaluation of the proposed method for phase reconstruction. (a) Measured relative receive phase between channel #4 and #15. (b) Calculated relative phase between the reconstructed absolute phase of channel #4 and #15.

Figure 8.

Reconstructed electrical property profiles along x-direction across the center of the phantom. The black solid lines are the probe measured value; the red dash lines indicate the reconstructed value.