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A feasibility study of magnetic resonance driven electrical impedance tomography using a phantom

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Abstract

Imaging the electrical properties of human tissue may aid in cancer diagnoses or monitoring organ function. Traditionally, the electrical properties are revealed with electrical impedance tomography, where currents are injected into human tissue and voltages are measured on the surface. This paper focuses on a method of measuring the electrical properties using a magnetic resonance (MR) scanner without current injection. In magnetic resonance driven electrical impedance tomography (MRDEIT), the MR phenomenon is used to induce currents in the body and the complex permittivity map is inversely computed from the difference between the modeled electric field and the actual surface electrode measurements. Computer simulations indicate that with noise level under 20%, the contrast is visually discernible in the reconstruction image. A phantom experiment is demonstrated and this supports results from computer simulation studies. The noise level in electrode measurements is evaluated to be approximately 7.8% from repeated experiments, confirming the potential to reconstruct conductivity contrast using MRDEIT. With further improvements in hardware and image reconstruction, MRDEIT may provide an additional contrast mechanism reflecting the electrical properties of human tissue, which may ultimately be used to diagnose a cancer or assist in electroencephalography.

Keywords: electrical impedance tomography, magnetic resonance, MR driven EIT

(Some figures may appear in colour only in the online journal)

1. Introduction

Electrical impedance tomography (EIT) is a technique that allows the mapping of the electrical properties (conductivity and permittivity) of tissue from surface electrode measurements. Tissue electrical properties are associated with intra- and extra-cellular ion concentrations (conductivity) and the capacity of bi-lipid cell membranes (permittivity). Different tissues
may exhibit different electrical properties, which can be used to identify tissue types. Studies have reported that significant conductivity or permittivity differences exist between abnormal and benign tissues in the breast (Da Silva et al 2000), prostate (Halter et al 2009), cervix (Brown et al 2000) and skin (Aberg et al 2004). In the biomedical imaging field, EIT has been studied in a number of applications including lung functioning (Adler et al 2009), breast cancer detection (Halter et al 2008) and brain function measurements (Romsauerova et al 2006). Some of these applications have been commercialized and are clinically used to monitor physiological activities (e.g. Dräger PulmoVista® 500®).

The merits of EIT include low equipment and operation costs, the non-invasive nature of the methods and high temporal resolution (Holder 2004). However, the spatial resolution of EIT is not comparable to CT or MRI. Unlike these conventional imaging modalities where images are directly computed from measurement data, EIT image reconstruction is based on solving an inverse problem. The reconstruction usually involves a finite element model, which computes the current field and electrode measurements based on an assumed electrical property distribution and excitations (forward problem). Electrical property changes will lead to corresponding measurement changes. Inversely, the electrical property changes can be inferred from these measurement changes, and this is the inverse problem in the EIT image reconstruction. Because of the nonlinear, diffusing nature of the current field, the inverse problem is severely ill-posed. Recovering the electrode properties is vulnerable to small measurement errors (Uhlmann 2009).

Absolute and difference imaging are two types of methods in EIT. In absolute imaging, the EIT image is reconstructed from only a single set of measurement data. In difference imaging, multiple sets of measurement data are collected to compute the change in electrical properties. One data set is considered as the baseline, and the differences between all other measurements and this baseline are used in the reconstruction algorithm. Compared to absolute imaging, difference imaging eliminates systematic errors in the measurements (e.g. calibration errors) and often yields images with higher accuracy. Difference imaging can also be utilized in cancer detection, with baseline data acquired by imaging a phantom of homogenous electrical properties (Wan 2012).

Recently, increasing attention has been focused on acquiring the tissue electrical properties with magnetic resonance (MR) imaging. MR electrical impedance tomography (MREIT: Zhang 1992, Woo et al 1994, Ider and Birgul 1998) combines EIT and current density imaging (Joy et al 1989, Scott et al 1991) to achieve a high spatial resolution. MREIT applies low frequency (a few kilohertz) current into the imaging object through surface electrodes and measures the effect of the applied current on the magnetic flux density using the MR scanner. Experimental phantom and human studies (Oh et al 2005, Kim et al 2009) have demonstrated the feasibility of imaging with MREIT. A recent study (Sadleir et al 2013) demonstrated that using currents within the human safety limits, the contrast is achievable although with compromised signal to noise ratio (SNR). A new approach that avoids the need to inject current is electrical property tomography (EPT). Recent phantom studies have been reported (Katscher et al 2009, Voigt et al 2011).

Magnetic resonance driven electrical impedance tomography (MRDEIT, Negishi et al 2011) is an alternative approach to image the conductivity and permittivity with MR. The MR phenomena itself ($M_0$) from individual voxels is used as the excitation source, thus eliminating the need for current injection as in traditional EIT. The rotating magnetization creates a time-varying electromagnetic (EM) field outside the excitation voxel and results in voltage measurements on electrodes attached to the imaged object’s surface. The conductivity

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1 Dräger PulmoVista® 500. KGaA, Drägerwerk AG & Co. 53–55, Moislinger Allee 23558 (Lübeck, Germany).
A feasibility study of MRDEIT using a phantom and permittivity map may be computed by solving an inverse problem, which minimizes the difference between modeled measurements and actual measurements. MRDEIT provides an additional contrast mechanism reflecting the electrical properties to the existing MR image information and may be potentially used to detect tumors, assess local specific absorbing rate (SAR) of radio frequency (RF) pulse in MR scanning and improve source localization from electroencephalographic and magnetoencephalographic data. The work presented here assesses the feasibility of the MRDEIT method and presents initial findings based on a phantom study.

2. Methods

2.1. Mathematical formulation

2.1.1. Signal source. In contrast to traditional EIT applications where electrical currents are injected into the imaging object through surface electrodes, the MRDEIT signal is generated from the magnetization $M_0$ rotating in the transverse plane perpendicular to the $B_0(z)$ direction at echo time ($TE$) after the 90° RF excitation pulse during a routine MR sequence (spin echo or gradient echo). The magnetization $M_0$ is the net summation of numerous precessing nuclear spins and each nuclear spin can be considered as a magnetic dipole with magnetic moment $m$. The magnetic moment $m$ results in the magnetic flux intensity $B(m)$ outside the magnetic dipole (Chow 2006):

$$B(m) = \frac{\mu}{4\pi} \left( \frac{3r(m \cdot r)}{r^5} - \frac{m}{r^3} \right),$$

where $\mu$ is permeability of material and $r$ is the position vector from the magnetic dipole. The rotating magnetic field $B$ created by the rotating magnetization $M_0$ in the transverse plane is the summation of $B(m)$ from individual magnetic dipoles:

$$B = \sum_{k=1}^{N} B(m_k),$$

$N$ being the number of magnetic dipoles in the exciting voxel. This rotating magnetic field $B$ is the source of MRDEIT signals.

2.1.2. Forward problem. Based on electrical properties, solving the forward problem starts with computing the vector electric field distribution $E$ in the imaging domain. According to Faraday’s law of induction, the rotating magnetic field $B$ induces a changing electric field $E$:

$$\nabla \times E = -\frac{\partial B}{\partial t}.$$  \hfill (3)

The induced electric field generates eddy currents if a conductor exists in the electric field. Combining this with Ampere’s law:

$$\nabla \times B = \sqrt{\mu \varepsilon} \cdot \frac{\partial E}{\partial t},$$

where complex permittivity $\varepsilon = \varepsilon + i\sigma \omega$ consists of permittivity $\varepsilon$ and conductivity $\sigma$, and $\omega$ is angular frequency (the Larmor frequency corresponding to $B_0$ in MRDEIT).

Combining (3) and (4), we obtain Maxwell’s wave equation:

$$\nabla \times \nabla \times E - k^2 E = 0,$$

where $k = \sqrt{\mu \varepsilon \omega}$ is the wavenumber of the electromagnetic wave. The double curl operation $'\nabla \times \nabla \times'$ in (5) is not approximated with $'\nabla^2'$ as in traditional electromagnetic wave
equations because the assumption of instant charge redistribution is not made. The electric field \( \mathbf{E} \) induces voltage potentials that are recorded by the electrodes, which are the measurements in MRDEIT:

\[
V = V_{\text{gnd}} + \int_C \mathbf{E} \cdot dl,
\]

where \( V_{\text{gnd}} \) is the reference level for the potential function and the integral of electric field intensity designates the electrode voltage above the reference ground.

2.1.3. Inverse problem. In order to compute the complex permittivity inside the imaging domain from voltage measurements on the surface electrodes, a model is employed to simulate the electric field in the imaging domain based on excitation and complex permittivity distribution (forward problem). This model is similar to that in traditional EIT image reconstruction, except that the field is represented by a vector field (electric field intensity \( \mathbf{E} \)) in MRDEIT as opposed to a scalar electric potential field \( \psi \) in EIT. The MRDEIT image is reconstructed by computing the complex permittivity values in the models so that the model-based measurement simulation matches the real world measurements (inverse problem).

The model first assumes a homogeneous complex permittivity \( \hat{\varepsilon}_{\text{baseline}} \) in the imaging domain and its corresponding voltage measurements \( V_{\text{baseline}} \) on the electrodes. A Jacobian matrix \( J \) is defined to depict the change in voltage measurements with respect to complex permittivity changes:

\[
J = \begin{pmatrix}
\frac{\partial V_1}{\partial \hat{\varepsilon}_{1}} & \cdots & \frac{\partial V_1}{\partial \hat{\varepsilon}_{N}} \\
\vdots & \ddots & \vdots \\
\frac{\partial V_L}{\partial \hat{\varepsilon}_{1}} & \cdots & \frac{\partial V_L}{\partial \hat{\varepsilon}_{N}}
\end{pmatrix},
\]

where \( V_l (l = 1, 2, \ldots, L) \) is the voltage measurement on electrode \( k \) in a total of \( L \) electrodes. The imaging domain is discretized into \( N \) small regions, and \( \hat{\varepsilon}_n (n = 1, 2, \ldots, N) \) designates the complex permittivity in region \( n \).

A difference imaging method is employed in the MRDEIT reconstruction. Small deviations in the complex permittivity \( \delta \varepsilon \) from a homogeneous baseline can be computed inversely from the measurement voltage difference \( \delta V \) between a contrast condition (inhomogeneous complex permittivity) and the baseline condition. In a phantom or computer simulation study, the contrast condition refers to the condition where the imaging domain has complex permittivity contrasts, while the baseline condition refers to the condition where it is homogeneous. For \textit{in vivo} breast cancer detection for example, the contrast condition data are the actual \textit{in vivo} data acquired from the subject’s breast immersed in a saline bath that has conductivity similar to that of the breast, and the baseline condition data are obtained from imaging a homogeneous saline bath. The inverse problem of recovering complex permittivities is severely ill-posed (Holder 2004) in the sense that small boundary measurement errors can lead to large reconstructed permittivity variations within the imaging domain. Tikhonov regularization, a common technique used in solving inverse problems, is employed to reduce the effect of measurement noise on image reconstruction by imposing a continuity of complex permittivity across the imaging domain (Holder 2004):

\[
\delta \varepsilon = (J^*J + \lambda^2 L^*L)^{-1} J^* \delta V,
\]

where superscript * and \(^{-1}\) designate the conjugate transpose and inverse of a matrix, respectively. \( \lambda \) is the Tikhonov parameter and \( L \) is the matrix form of partial differential operator (e.g. Laplacian) and it is approximated with an identity matrix \( I \) in this study. While both the permittivity and conductivity can be obtained from the reconstructed complex permittivity, the conductivity image (imaginary part) is the primary focus of this study.
2.1.4. Data processing for MR images. The MRDEIT measurement data are acquired with electrodes attached to the surface of the imaging phantom and their outputs are connected to the conventional MR data acquisition system. In this mode electrodes can be used to either replace the conventional receiver coil (as in the experiments described below) or complement the conventional receiver coils. Standard MR images can be reconstructed from these electrode measurements since the potentials recorded are in fact phase and frequency encoded using conventional MR imaging sequences. Images are reconstructed as in conventional MR imaging with a 2D Fourier transform, after which each pixel intensity in the MR represents the complex voltage signal (with magnitude and phase) sensed on the electrodes. Assuming that the $T_1$ and $T_2$ effects are negligible due to the use of a long TR and a short TE, the complex signal can be expressed as (modified from Hoult (2000)):

$$S_a(r) = V_{dr} \cdot E_a(r) \cdot M_0(r) \cdot \sin(V_T |H^+(r)|\alpha) \cdot e^{i\phi^+(r)}$$

where $\alpha$ is the nominal flip angle, $H^+(r) = |H^+(r)| e^{i\phi^+(r)}$ is the resulting transmitting magnetic field intensity at the excitation voxel $r$ from a unit current on the RF coil, $V_T$ is related to the RF transmitter sensitivity, $M_0(r)$ is the magnetization magnitude related to proton density, $V_{dr}$ is associated with the nth sensor (electrode) configuration and $E_a(r)$ is the complex receiving electric field intensity (scalar) at the sensor created by a unit excitation at the voxel. Both $H^+$ and $E_a(r)$ are dependent on the electrical properties across the imaging field, and the latter is used to compute the complex permittivity map in MRDEIT image reconstruction.

The purpose of preprocessing the MR image in MRDEIT is to extract $E_a$ from the signal in the presence of non-uniform excitations resulting from inhomogeneous $|H^+|$ field and variations in spin density $M_0$ (9). $|H^+|$ (or $B_1^+$ mapping) is obtained with the double angle method in this study (Wang et al 2005). To be specific, additional standard body coil images are collected with flip angles of $\alpha$ and $\alpha/2$ for both baseline condition (homogeneous complex permittivity) and contrast condition, respectively:

$$S_a(r) = V_{bc} \cdot H^-(r) \cdot M_0(r) \cdot \sin(V_T |H^+(r)|\alpha) \cdot e^{i\phi^+(r)}$$

$$S_x(r) = V_{bc} \cdot H^-(r) \cdot M_0(r) \cdot \sin\left(V_T |H^+(r)|\frac{\alpha}{2}\right) \cdot e^{i\phi^+(r)}.$$

The signal equations are similar to (9) except that the sensor sensitivity $V_{dr}$ is replaced by $V_{bc}$ for body coil sensitivity, and the receiving electric field intensity $E_a(r)$ is replaced by receiving magnetic field intensity $H^-(r)$. $|H^+(r)|$ is determined by

$$|H^+(r)| = \frac{2}{V_T \cdot \alpha} \cos^{-1}\left(\frac{S_a(r)}{2S_x(r)}\right).$$

Under the assumption that the transmitting and receiving magnetic fields are identical (Katscher et al 2009), the signal equation for body coil image is

$$S_a(r) = V_{bc} \cdot |H^+(r)| \cdot M_0(r) \cdot \sin(V_T |H^+(r)|\alpha) \cdot e^{i2\phi^+(r)}.$$

It is challenging to determine the absolute phase $\phi^+(r)$ of the transmitting field (Katscher et al 2009). However, the phase difference $(\phi^+_b(r) - \phi^+_c(r))$ between the two acquisitions can be calculated by taking the ratio of the body coil images:

$$e^{i2(\phi^+_b(r) - \phi^+_c(r))} = \frac{S_{abc}(r)}{S_{ab}(r)} \cdot \frac{|H^+_{bc}(r)|}{|H^+_{ab}(r)|} \cdot \frac{\sin(V_T |H^+_{bc}(r)|\alpha)}{\sin(V_T |H^+_{ab}(r)|\alpha)}.$$

The subscripts $b$ and $c$ designate the baseline and contrast acquisitions, respectively. To eliminate the effect of spin density $M_0(r)$ variations, the complex ratio of the electric field $E_a(r)$ in the contrast condition to the baseline conditions is computed:
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Enc(r) Enb(r) = Snb(r) · sin(VT |Hb(r)|α) sin(VT |Hb(r)|α) · e^{-i(φb(r)−φb(r))}.

(15)

To summarize, the electric field intensity ratio in this study is obtained with a total of six data acquisitions: two MR body coil acquisitions with different flip angles in baseline condition, the same two MR body coil acquisitions in contrast condition and two electrode measurements in baseline and contrast conditions. The ratio of the electric field in (15) contains the measurement difference and is used to compute the MRDEIT reconstruction.

2.2. Modeling

The entire computation space for the phantom experiment is modeled with a 3D finite element mesh consisting of 25 × 25 × 25 linear brick elements. A complex scalar is assigned to each element to represent the electrical property (complex conductivity) within that element.

The electric field in the model is determined by vectors defined on all the element edges. Specifically, given the electric field vectors on 12 edges of a single element, the electrical field within the element can be computed with the interpolation equation

\[ E_e = \sum_{i=1}^{12} N_e^i E_{e}^i, \]

(16)

where \( N_e^i \) is vector basis function and \( E_{e}^i \) is the electric field intensity defined on edge \( i \) in the element \( e \) (Jin 2002). The interpolation function is used to ensure the electric field continuity in the entire computation space.

A right-hand Cartesian coordinate system is employed to describe the model and compute the field intensities in the mesh. Specifically, the \( z \)-axis is aligned with the main magnetic field (\( B_0 \)) of the MR scanner, the \( y \)-axis is vertically pointing downward and the \( x \)-axis is perpendicular to both \( y \)- and \( z \)-axes (figure 1).

The computation space is divided into several sub-regions in order to match the phantom experiment configurations (figure 1). Specifically, a mesh grid of 15 × 15 × 15 elements located at the center of the computation space designates the imaging domain, modeling the saline bath in the phantom experiment. One layer of elements immediately outside the imaging domain is the imaging boundary, simulating the acrylic sides of the saline bath container. Four square electrodes consisting of nine elements (3 × 3) are attached to both lateral outer surfaces of the imaging boundary. The space outside the imaging boundary is primarily filled with elements of the same electrical property as air. ‘Wire’ elements extend outward from the centers of the four electrodes. On each of the four ‘wires’, an element is defined as ‘output’ with an impedance equal 50 Ω, matching the input impedance of the coaxial cable. These ‘output’ elements are employed to model the voltage measurements at the end of the coaxial cables when an element in the imaging domain is excited. The outermost layer of elements in the computation domain is modeled as a metal shell ground, where a second-order absorbing boundary condition (Volakis et al 1998) is imposed to ensure zero reflecting electric field. The electrode voltage measurements are referenced to this metal ground.

Physically, all the voxels in an imaging slice are excited at the same time by a standard MR sequence and the measurements represent the summation of the signals resulting from all the excitations. However, because each excitation is uniquely coded with frequency and phase, signals from individual excitations can be identified distinctly. In the computational modeling, voxels are mathematically treated as if they are excited individually—the magnetization \( M_0 \) is set to zero throughout the computation space except for the excited voxel (element in the mesh). Since \( M_0 \) is rotating in the transverse plane perpendicular to the \( B_0 \) direction (\( z \)-axis), the excitation is modeled as a complex vector with real and imaginary parts being unit vectors
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Figure 1. (a) 3D MRDEIT model with defined elemental regions. The coordinate system is defined as shown in the upper left corner. (b) Cross-section of the MRDEIT model at the electrode level ($y = 0.10$). (c) Cross-sections of the MRDEIT model at the levels above and below the electrode plane ($y = 0.15$ and $y = 0.05$).

on the four $x$-direction and four $y$-direction edges of the excited element, respectively. The electric field is established in the computation space after the excitation, and voltage outputs
Table 1. Electrical properties defined in different regions in the mesh.

<table>
<thead>
<tr>
<th>Region</th>
<th>Conductivity ($\text{S m}^{-1}$)</th>
<th>Permittivity ($\times 8.85 \times 10^{-12} \text{ F m}^{-1}$)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Imaging domain background</td>
<td>1.8$^a$</td>
<td>77$^a$</td>
</tr>
<tr>
<td>Imaging domain boundary</td>
<td>0</td>
<td>2</td>
</tr>
<tr>
<td>Electrodes, wire and outermost metal shell</td>
<td>6.0e7$^b$</td>
<td>1</td>
</tr>
<tr>
<td>Output element</td>
<td>2.31$^c$</td>
<td>0</td>
</tr>
<tr>
<td>Lesion</td>
<td>3.4$^a$</td>
<td>75$^a$</td>
</tr>
<tr>
<td>Air</td>
<td>0</td>
<td>1</td>
</tr>
</tbody>
</table>

$^b$ Griffiths (1999).
$^c$ Computed from 50 $\Omega$ coaxial cable.

at the electrodes are computed from the electric field vectors on the $x$-direction edge in the output elements (6).

For each excitation, the forward model computes electric field intensities on a total of 50,700 edges. In the inverse computation, a total of 3375 unknown complex permittivities (number of hexahedron elements $15 \times 15 \times 15$ in the imaging domain) are obtained by solving a system of equations whose number is equal to the number of excitations times the number of electrodes $15 \times 15 \times 15 \times 4$.

2.3. Simulation study

In the simulation study, the baseline condition is established by assigning electrical conductivity and permittivity values to the elements in the mesh regions according to table 1. A cylindrical ‘lesion’ is positioned in the imaging domain to simulate the abnormality (contrast condition) from baseline. The ‘lesion’ has a diameter of three elements and the height of the imaging domain. For contrast, the conductivity in the ‘lesion’ is chosen to be twice the value in the baseline (a contrast of 100%). The simulation of the contrast condition is demonstrated in figure 2.

The simulated measurement voltages are computed for individual excitations under both the baseline and contrast conditions using the forward model. The output voltage differences from the two conditions are used to compute the complex permittivity changes in the contrast condition relative to the baseline condition with the inverse solver. During this process, various levels of noise are added into the voltage differences to assess the inverse solver’s robustness to measurement errors. In this simulation study, the additive noise is normally distributed with a mean 0 and standard deviation $\sigma_{\text{noise}}$, where

$$\sigma_{\text{noise}} = L_{\text{noise}} \cdot \max(|V_c - V_b|).$$  \(17\)

In (17), the voltage outputs in baseline and contrast conditions are designated by $V_b$ and $V_c$, respectively. The levels of noise $L_{\text{noise}}$ tested in this study are 0%, 1%, 2%, 5%, 10% and 20%. In the simulation, the MRDEIT images are reconstructed with a Tikhonov regularization parameter $\lambda = 0.01$ chosen empirically.

2.4. Phantom study

The MRDEIT phantom comprises a five-sided acrylic cube measuring 12 cm $\times$ 12 cm $\times$ 12 cm and a cylindrical glass bottle 4.2 cm in diameter positioned
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Figure 2. A ‘lesion’ is positioned in the imaging domain of the mesh to simulate the contrast condition.

in the cube (figure 3). Two copper electrodes (2.4 cm × 2.4 cm) are attached to two lateral surfaces of the cube (four electrodes in total), with locations matching those defined in the mesh model. The electrodes are cut into a ‘star’ shape in order to reduce eddy currents induced on the electrode surface (figure 4). The electrodes are connected to a custom-designed preamplifier unit by coaxial cables with inner conductors providing pathways for electrical signals and outer conductors as paths to ground. The amplified electrode voltage signals are measured using a standard MRI data acquisition scheme (see details below).

For the baseline condition, the acrylic cube and bottle are both filled with 1.2% sodium chloride (NaCl), 4 mM L$^{-1}$ copper (II) sulfate (CuSO$_4$) and deionized water solution (by weight). The conductivity of the solution is approximately 1.8 S m$^{-1}$ according to Peyman et al (2007). Voltage measurements from the four electrodes, normal angle ($\pi/2$) and half flip angle ($\pi/4$) body coil measurements are conducted consecutively using a gradient-echo sequence (TR/TE = 3000 5 ms$^{-1}$, FA = 90, 25 5 mm-slices, 128 × 128 matrix, FOV 15 cm × 15 cm) on a 3 T MR scanner (Siemens TIM Trio, Erlangen, Germany). Each scan is completed in 8.5 min with 2 averages (NEX).

For the contrast condition, the lesion was created by using 2.4% sodium chloride in the glass bottle, which simulates a potential abnormal lesion in the object. After the sodium chloride is fully resolved in the solution, electrode voltage measurements with normal flip angle ($\pi/2$) and half flip angle ($\pi/4$) body coil measurements are sequentially conducted again using the same sequence parameters as in the baseline data collection.

Both the normal flip angle ($\pi/2$) and half flip angle ($\pi/4$) body coil measurement data are used to compute the $B_1^+$ mapping with (12) and eliminate its effect on the electrode measurement image. The measurement differences between the baseline and contrast condition are used to inversely compute the changes in electrical properties in the imaging domain. A Tikhonov regularization parameter $\lambda = 0.1$ was used in the image reconstruction.
3. Results

3.1. Forward model

3.1.1. Simulated electric field distribution. For the baseline condition, the electric field distribution is simulated by modeling a unit single voxel excitation at the center of the imaging domain (0.104, 0.104, 0.104). The resulting electric field vectors in the vicinity of this voxel are shown in figure 5 and the magnitude of the electric field for the entire modeled space is shown in figure 6. Both the real and imaginary components of the electric field rotate about the excitation field. The magnitude of the electric field attenuates with the distance away from the excitation. The electric field intensity at the electrodes is approximately $3.6 \times 10^{-5}$ V m$^{-1}$, or 3% of the maximum electric field (at the excitation) in the mesh model.

3.1.2. Simulated measurements versus experiment measurements. The electrode voltage measurements are simulated for every single voxel (pixel in 2D cross-section images) excitation and compared with the real phantom experiment measurements. Specifically, the magnitude of voltage measurements is represented by pixel intensity in the coronal electrode MR images for each excitation at the particular pixel. Similarly, in the simulated measurement images (2D $xz$ cross-sections of the 3D imaging domain along the vertical direction), the grayscale intensity of each pixel indicates the voltage magnitude sensed on the electrode resulting from excitation at the pixel. The spatial pattern in the simulated measurement images (figure 7(a)) is similar to the actual measurements (MR electrode images, figure 7(b)) in the sense that the voltage measurements are greater (bright) on both sides of the electrode, while the resulting
Figure 4. A photograph of the copper electrode. The electrode is cut into a ‘star’ shape to reduce eddy current effects on the metal surface during data acquisition.

Figure 5. The real and imaginary parts of an electric field created by an excitation at the center of the mesh model.

voltage is approximately zero (dark) in regions close to the electrode, the imaging domain center, and the opposite side of the sensing electrode. A similar pattern exists in simulation and MR image pairs from the other three electrodes.
3.2. Image reconstruction

3.2.1. Simulations: effect of different levels of noise. The 3D reconstructed conductivity difference image (imaginary part of complex permittivity) without added noise is shown in figure 8(a). The corresponding coronal 2D cross-section slices are shown in figure 8(b). The mean conductivity difference in the ‘lesion’ is 1.07 S m$^{-1}$ with a maximum value of 1.71 S m$^{-1}$ at the lesion center, compared to $3.87 \times 10^{-4}$ S m$^{-1}$ in the background bath.

The conductivity images reconstructed under different simulated noise levels are shown in figure 9. Increasing levels of noise lead to deteriorating reconstruction quality with growing conductivity variation in the background. For noise levels exceeding 10%, it is difficult to distinguish the lesion from its surroundings.

The effect of additive noise in the difference measurement on the reconstruction is quantitatively evaluated with contrast to noise ratio (CNR) analysis. Contrast is defined as the mean conductivity difference of the lesion region to the background region and noise is the standard deviation of conductivities in the background. The CNRs for different noise levels are tabulated in table 2. The CNR drops from 17.75 to 3.5 dB as noise level increases from 0 to 20%.

3.2.2. Phantom experiment. The $|H^+|$ is inhomogeneous across the imaging field mainly due to variations of the main magnetic field and inhomogeneity induced by the imaging object. Specifically, the $|H^+|$ intensity (B$_1$ map) of the center slice for baseline condition is shown in figure 10. This inhomogeneity effect is corrected by (15) to ensure that the corrected phantom measurements are the result of identical excitations across all the voxels, matching the unity sources in simulated forward model. This is necessary for solving the model-based inverse problem.
Figure 7. (a) Simulated measurement images from electrode 1 for excitations at every individual pixel. (b) Actual electrode MR images from electrode 1. Dark circles denote glass bottle boundaries.

After processing the MR data as specified in section 2.1.4, the measurement difference between contrast and baseline conditions is used to compute the conductivity image. The difference data on electrode 2 (closest to contrast) for the phantom experiment and the
**Figure 8.** (a) 3D conductivity reconstruction image without simulated noise. Dotted planes indicate orientation and order of 2D slices in (b). Electrode locations are also demonstrated. (b) 2D cross-section slices of image (a) along the y-axis. Color bars in both images represent reconstructed conductivity in S m$^{-1}$.

Simulation are illustrated in figure 11. Despite the artifacts on the phantom boundary and the noise, the measurement difference data are similar to the simulation in the sense that the pixels on both sides of the contrast have higher intensity than the background because the elevated conductivity mostly affects the signals generated from these pixels.
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Figure 9. Reconstructed conductivity images with various additive measurement noise (0 to 20%). The images are 2D coronal cross-sections at the level of the electrode center (same for all four electrodes).

Table 2. Effect of added noise on simulated reconstruction image.

<table>
<thead>
<tr>
<th>Noise level</th>
<th>Mean conductivity in the lesion ($S\cdot m^{-1}$)</th>
<th>Mean background conductivity ($S\cdot m^{-1}$)</th>
<th>Background conductivity standard deviation ($S\cdot m^{-1}$)</th>
<th>Contrast to noise ratio (dB)</th>
</tr>
</thead>
<tbody>
<tr>
<td>0</td>
<td>1.07</td>
<td>$3.87 \times 10^{-4}$</td>
<td>$1.89 \times 10^{-2}$</td>
<td>17.75</td>
</tr>
<tr>
<td>1%</td>
<td>1.07</td>
<td>$1.4 \times 10^{-3}$</td>
<td>$2.99 \times 10^{-2}$</td>
<td>15.52</td>
</tr>
<tr>
<td>2%</td>
<td>1.06</td>
<td>$2.2 \times 10^{-3}$</td>
<td>$5.06 \times 10^{-2}$</td>
<td>13.21</td>
</tr>
<tr>
<td>5%</td>
<td>1.05</td>
<td>$2.7 \times 10^{-3}$</td>
<td>$1.22 \times 10^{-1}$</td>
<td>9.39</td>
</tr>
<tr>
<td>10%</td>
<td>1.04</td>
<td>$3.6 \times 10^{-3}$</td>
<td>$2.41 \times 10^{-1}$</td>
<td>6.33</td>
</tr>
<tr>
<td>20%</td>
<td>1.13</td>
<td>$8.8 \times 10^{-3}$</td>
<td>5.03</td>
<td>3.50</td>
</tr>
</tbody>
</table>

The conductivity magnitude image reconstructed from the phantom data is demonstrated in figure 12. The actual location of the lesion is determined by its boundary from MR body coil image. The lesion boundary is superimposed on the reconstructed images and indicated by a blue circle. The lesion consists of a number of elevated conductivity pixels (red) from the background (primarily green). The mean reconstructed conductivities within the lesion and the background image for all 15 cross-sections are shown in table 3. For the 3D volume, the average reconstructed conductivity is 0.0998 and 0.0210 $S\cdot m^{-1}$ in the lesion and background region, respectively.

The location of the lesion in the reconstructed images is also compared with its actual location specified by the MR body coil image. The position in each slice of the lesion center is computed by averaging the coordinates of the pixels whose values are greater than a threshold, which is half the maximum conductivity in the slice (artifact pixels on the peripheral are omitted). The deviations of the reconstructed lesion center from the actual center in all the reconstructed cross-sections are quantified in table 3. The average spatial error in the phantom reconstruction is 5.80 mm across all 15 slices where the FEM model mesh size is 8.7 mm.
Figure 10. $B_1$ map showing the inhomogeneous $|H_{\parallel}|$ field in the phantom experiment due to variations of main magnetic field and imaging object induced inhomogeneity.

Figure 11. The measurement difference on electrode 2 (closest to the contrast) from the phantom experiment is compared with that from the simulation.

4. Discussion

This paper presents a first phantom study to demonstrate the feasibility of a method to compute the electrical property of an imaging field based on surface electrode measurements resulting from the magnetic resonance phenomena in standard MR scanning. The reconstructed image (figure 12) shows that changing the conductivity in part of the phantom provided detectable contrast in the image. We also demonstrated that in the simulation, for noise level of less than 10%, the contrast is discernible with the MRDEIT method. A number of additional phantom experiments were conducted to evaluate the SNR for the real electrode measurements. The
Figure 12. Reconstructed conductivity magnitude images (coronal cross-section) of the phantom experiment. In each reconstructed cross-section image, the lesion boundary is designated with a blue circle, the location of which is determined by the body coil MR image (lower right).

Table 3. Mean conductivity in the lesion and background regions and the lesion center location error for each slice in the MRDEIT reconstruction from the phantom experiment.

<table>
<thead>
<tr>
<th>Slice number</th>
<th>Mean conductivity in lesion (S m(^{-1}))</th>
<th>Mean conductivity in background region (S m(^{-1}))</th>
<th>Location error of reconstructed lesion center (mm)</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>0.0670</td>
<td>0.0398</td>
<td>17.5</td>
</tr>
<tr>
<td>2</td>
<td>0.0934</td>
<td>0.0271</td>
<td>6.98</td>
</tr>
<tr>
<td>3</td>
<td>0.1154</td>
<td>0.0284</td>
<td>4.34</td>
</tr>
<tr>
<td>4</td>
<td>0.1223</td>
<td>0.0278</td>
<td>3.58</td>
</tr>
<tr>
<td>5</td>
<td>0.1240</td>
<td>0.0256</td>
<td>2.17</td>
</tr>
<tr>
<td>6</td>
<td>0.1294</td>
<td>0.0228</td>
<td>2.82</td>
</tr>
<tr>
<td>7</td>
<td>0.1337</td>
<td>0.0214</td>
<td>2.44</td>
</tr>
<tr>
<td>8</td>
<td>0.1345</td>
<td>0.0203</td>
<td>4.09</td>
</tr>
<tr>
<td>9</td>
<td>0.1292</td>
<td>0.0183</td>
<td>5.91</td>
</tr>
<tr>
<td>10</td>
<td>0.1186</td>
<td>0.0166</td>
<td>5.91</td>
</tr>
<tr>
<td>11</td>
<td>0.1066</td>
<td>0.0144</td>
<td>6.28</td>
</tr>
<tr>
<td>12</td>
<td>0.0915</td>
<td>0.0124</td>
<td>5.91</td>
</tr>
<tr>
<td>13</td>
<td>0.0621</td>
<td>0.0110</td>
<td>8.58</td>
</tr>
<tr>
<td>14</td>
<td>0.0252</td>
<td>0.0119</td>
<td>5.96</td>
</tr>
<tr>
<td>15</td>
<td>0.0439</td>
<td>0.0169</td>
<td>4.10</td>
</tr>
</tbody>
</table>
noise level (the ratio of standard deviation to the maximum measurement intensity, as defined in the simulation) is found to be approximately 7.8%, which falls within the contrast-discernible noise range. We expect the ultimate sensitivity for a 100% contrast would be similar to the reconstructed images with 5%–10% noise level in figure 9. Although not perfect, the result of this initial phantom study demonstrates that it is promising to use MRDEIT for detecting conductivity contrasts. A complete sensitivity study of the MRDEIT method is necessary to estimate its ability to image objects of different sizes and various contrasts.

Currently, difference imaging is employed in MRDEIT. Baseline condition may be obtained by imaging a homogeneous saline bath in order to assess in vivo electrical properties clinically. We are also working on developing an absolute imaging method, where imaging the baseline condition is not required. With the MRDEIT approach, an alternative property contrast (complex permittivity) may be interrogated in addition to the traditional MR parameters (proton density, $T_1$, $T_2$, and so on) without extensive imaging system modifications or prolonging the data acquisition time. The computed complex permittivity may potentially be used to differentiate tissue types (cancer versus benign tissue), or to calculate local SAR.

The simulation study demonstrates that the measurements simulated using the forward model have a pattern similar to the actual electrode measurements (figure 7): the voltage generated by the excitation directly in front of the electrode is small while larger voltages are created by excitations close to the periphery of the electrodes. This may result from the fact that excitations are creating EM fields in a rotating manner (figure 5). The electric fields are tangential to the electrode surface from excitations in front of the electrodes so the sensed perpendicular electric field (measurements) is small. On the other hand, the excitation close to the periphery of the electrodes will generate perpendicular electric fields on the electrodes, resulting in greater voltage measurements.

Using the glass bottle is not a perfect solution for holding the contrast solution. However, although glass is a good insulating material with a very low conductivity, at a radio frequency of 123 MHz the EM wave can still travel through the glass wall without too much attenuation. In other words, the glass may be considered as a capacitor in circuit theory, which has low impedance on the transmitted signal, although it does change the phase. The point is that, unlike in a low frequency case, the conductivity variation within the bottle (contrast) will lead to an output signal change through a dielectric link (glass). In addition, since a difference imaging scheme is used in the phantom experiment, the effect of the glass interface is eliminated by comparing the signals from the contrast condition to that from the baseline condition. Conductive material (carbon-fiber, metal needle on the glass wall) for the container will increase the contrast signal strength, but the induced eddy currents in the conductors and susceptibility inhomogeneity may impair the received signals. We are currently investigating an alternative electrical translucent membrane (natural membrane from lamb’s intestine) for holding the contrast fluid, which may provide an improved interface (a lot thinner and the conductivity of which is closer to the background) between the background and the contrast.

The MRDEIT development is still in an early stage and faces many challenges. The average lesion center localization error is approximately 5.80 mm, which is minimal considering the mesh size of 8.7 mm used in the reconstruction. There is a moderate gap artifact of lower conductivity at the lesion center in the reconstructed image. The source of this artifact is still unknown. It may result from the model–measurement mismatch (dimension and/or position inaccuracy), approximation of $H^-$ by $H^+$ in (13), measurement noise or inappropriate selection of partial differential operator $L$ in (8) for inverse problem solving. Minimizing or even eliminating these reconstruction artifacts in MRDEIT is the focus of our current research. Potential improvements include: (1) increasing the number of electrodes. Four electrodes are used in this study while current EIT systems usually employ 32 channels or 64 channels.
Increasing the number of sensors will provide more information on boundary measurements, thus reducing the severity of the ill-posed inverse problem. (2) Using a finer FEM mesh which better matches the experiment configuration. The current mesh consists of same-sized hexahedrons. This is not accurate for modeling the cylindrical glass bottle lesion. Further development of using tetrahedron elements may contribute to finer mesh and provide flexibility to model round-shaped objects. (3) Increasing the capacitance between the electrodes and the body by using thinner insulators (currently the 1/4 inch acrylic walls act as an insulator). (4) Improving the impedance match between the electrodes and the coaxial cables. These improvements are being explored and we hope to see an improvement in the reconstruction.

The reconstructed conductivity in the lesion region has a mean of 0.0998 S m\(^{-1}\) and a maximum of 0.2794 S m\(^{-1}\), which is different from simulated reconstruction with a mean of approximately 1 S m\(^{-1}\) and maximum of approximately 1.7 S m\(^{-1}\). The reason for the large disagreement is primarily due to the selection of the Tikhonov regularization parameter \(\lambda\). In the phantom experiment, a higher \(\lambda\) is chosen (0.1 as opposed to 0.01 in simulation) to compensate the measurement noise and to reconstruct a smoother conductivity image (smaller \(\lambda\) may induce speckle noise in the reconstruction as shown in figure 13). The dependence of reconstructed conductivity on the regularization parameter may result in incorrectness in conductivity value and this was observed in other EIT reconstructions (Wan et al 2010). Nevertheless, the conductivity contrast is still preserved in the reconstruction and this contrast information is exploitable in the diagnostic MRDEIT applications.

In the phantom study, we collected a total of six MR images, three for baseline condition and three for contrast. However, in a clinical configuration, the baseline (saline solution) image could be obtained prior to the patient scans. Additionally, two of the three images are acquired with body coil in order to determine the \(B_1^+\) map using the double angle method. Alternative \(B_1^+\) mapping methods (e.g. Bloch–Siegert shift (Sacolick et al 2010)) do not require the two
additional body coil scans. Small flip angles can also be utilized to reduce the scanning time (TR) at a cost of decreasing signal strength. We will explore these different accelerating methods in our future research.

Electrical property tomography (EPT) is another technique to compute the electric conductivity and permittivity using MR phenomena. It computes the images from conventional MR signals, without requiring electrodes for current injection or signal detection. There are two main differences between the operation principles of EPT and MRDEIT. First, EPT measures magnetic field intensity (as in standard MR imaging), whereas MRDEIT measures electric field intensity. Inhomogeneous conductivity and permittivity changes the direction of the magnetic field intensity according to Ampere’s circuital law, which is detected in EPT from the curl (Katscher et al 2009), the gradient (Voigt et al 2011) or the Laplacian (Bulumulla et al 2012) of the magnetic field intensity. By contrast, conductivity and permittivity are computed from the electric field intensity at the electrodes in MRDEIT by an inverse operation but without taking any spatial gradients. Second, MRDEIT is essentially different from MREPT in the sense that MREPT computes the complex permittivity based on the local field intensities (neighboring voxels), while MRDEIT focuses on the global effect resulting from permittivity changes in the current pathways between the excited voxel and the electrodes. In other words, the complex conductivity in each element is determined by the measurements from all the excited voxels (each element is computed from over 13 000 \((15 \times 15 \times 15 \times 4)\) measurements in this study). Theoretically, the number of measurements could be over a million \((128 \times 128 \times 25 \times 4)\) if the FEM element size is comparable to voxel dimension in the MR image. The information to compute each complex permittivity value is several orders of magnitude larger than that in MREPT. However, the same global nature of the solution also introduces the inverse problem. Other disadvantages of MRDEIT include solutions largely dependent on measurement accuracy, similarities of model to the reality and the number of electrodes (boundary measurements). These factors limit the sensitivity and resolution in MRDEIT reconstruction images. With a finer mesh in the model matching the real experiment
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and more peripheral electrodes, we believe MRDEIT has the potential to compute complex permittivity distribution with higher accuracy and reconstruct images with higher resolution.

Figure 14 shows an EPT image obtained by applying equation (10) in Voigt et al (2011) to the $B_1^+$ data computed from the same data that was used to compute figure 12. Elevated conductivity changes are detected at the lesion location despite a moderate level of speckled noise across the images and the reconstructed conductivity change agrees with the actual phantom. Large errors are seen at the boundary of the lesion because of the discontinuity of the conductivity at the glass wall (Seo et al 2012). The noise in the image may be due to the use of the body coil in our experiment (instead of a head coil) and the use of 90° and 45° flip angles for $B_1^+$ mapping rather than standard 120° and 60°. The 90° excitation was preferred in MRDEIT for the maximum electric field intensity it generates.

5. Conclusion

This paper demonstrates the feasibility of imaging the electrical properties of tissue using the MRDEIT method through phantom studies. While some image artifacts are present, the reconstructed contrast object is visually discernible from the conductivity image. The MRDEIT approach is currently in its early developmental stage and may potentially provide an additional contrast mechanism in MRI reflecting the electrical properties of tissue, which may be exploited in cancer diagnosis, SAR assessment or source localization of brain activities in electroencephalography or magnetoencephalography.

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