Magnetic Resonance Electrical Impedance Tomography at 3 Tesla Field Strength

Suk H. Oh,1 Byung I. Lee,2 Tae S. Park,1 Soo Y. Lee,1,* Eung J. Woo,2 Min H. Cho,1 Jin K. Seo,3 and Ohin Kwon4

Magnetic resonance electrical impedance tomography (MREIT) is a recently developed imaging technique that combines MRI and electrical impedance tomography (EIT). In MREIT, cross-sectional electrical conductivity images are reconstructed from the internal magnetic field density data produced inside an electrically conducting object when an electrical current is injected into the object. In this work we present the results of electrical conductivity imaging experiments, and performance evaluations of MREIT in terms of noise characteristics and spatial resolution. The MREIT experiment was performed with a 3.0 Tesla MRI system on a phantom with an inhomogeneous conductivity distribution. We reconstructed the conductivity images in a 128 × 128 matrix format by applying the harmonic \( B_z \) algorithm to the \( z \)-component of the internal magnetic field density data. Since the harmonic \( B_z \) algorithm uses only a single component of the internal magnetic field data, it was not necessary to rotate the object in the MRI scan. The root mean squared (RMS) errors of the reconstructed images were between 11% and 35% when the injection current was 24 mA. Magn Reson Med 51:1292–1296, 2004. © 2004 Wiley-Liss, Inc.

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Information about electrical conductivity distribution inside biological tissues is useful for many purposes, such as modeling tissues to investigate action potential propagations, estimating therapeutic current distribution during electrical stimulation, and monitoring physiological functions (1–3). In combinatory studies of functional MRI (fMRI) with other brain mapping modalities, such as EEG and MEG brain mapping, information about conductivity distribution inside the brain is essential for the precise localization of activated regions (4,6). Electrical impedance tomography (EIT) is a conductivity imaging modality in which many surface electrodes are attached to an object to measure the electric potentials produced by injection currents. However, EIT has many drawbacks, including a limited amount of measured data, low sensitivity of the surface voltage to conductivity changes at the region far from the electrodes, and the ill-posedness of the corresponding inverse problem involved in image reconstruction. Recently, a combination of MRI and EIT (MREIT) was introduced as a new conductivity imaging modality (7–9). With MREIT, one can transform the ill-posed conductivity image reconstruction problem into a well-posed one by incorporating the internal data of magnetic flux density distributions measured by an MRI scanner. In previous MREIT experimental studies (10–12), the biggest problems were object rotations inside the magnet to acquire the three components of the internal magnetic field data, and low SNR of reconstructed images. However, new MREIT reconstruction algorithms were recently introduced, along with phantom imaging results obtained without any object rotations (13–15).

In this work we present some results of MREIT phantom imaging experiments performed with the harmonic \( B_z \) algorithm and a 3.0 Tesla MRI system. After briefly introducing the fundamental principle of the conductivity image reconstruction algorithm, we present the experimental results of the conductivity imaging performance evaluation.

MATERIALS AND METHODS

Conductivity Image Reconstruction Problem in MREIT

We assume a 3D electrically conducting object. Four electrodes are attached to the surface of the object, and two currents are then injected between two chosen pairs of electrodes. For example, \( I_1 \) is the injection current between a pair of electrodes facing each other, and \( I_2 \) is the injection current between the other pair. Then the voltages \( V_i \) and \( V_z \) inside the object due to the corresponding DC injection currents \( I_1 \) and \( I_2 \) satisfy the following partial differential equation:

\[
\begin{align*}
\nabla \cdot (\sigma(\mathbf{r})\nabla V_i(\mathbf{r})) &= 0 & \text{inside the object} \\
-\sigma \nabla V_i \cdot \mathbf{n} &= g_i & \text{on the surface}
\end{align*}
\]

[1]

where \( \sigma \) is the conductivity distribution inside the object, \( \mathbf{r} \) is a position vector, \( \mathbf{n} \) is the unit outward normal vector on the surface boundary, and \( g_i \) is the normal component of the current density due to the injection current \( I_i \). The current density \( \mathbf{J}_i \) inside the object is

\[
\mathbf{J}_i(\mathbf{r}) = -\sigma(\mathbf{r})\nabla V_i(\mathbf{r}) \quad \text{for } i = 1 \text{ and } 2.
\]

[2]

From Ampere’s law, the induced magnetic flux density \( \mathbf{B}_i \) is related to \( \mathbf{J}_i \) as...
In MREIT, we measure the induced magnetic flux density using an MRI scanner and try to reconstruct images of the conductivity distribution \( \sigma \) using the relations in Eqs. [1]–[3]. When the main magnetic field of the MRI scanner points to the \( z \)-direction, we can obtain images of \( B_i \) that is the \( z \)-component of \( B \) that is shown in Fig. 1. The magnetic field components \( B_i \) were determined to be 0.16 S/m and 0.255 S/m, respectively, by means of an impedance analyzer (4192A, Hewlett Packard). To evaluate the spatial resolution of the conductivity imaging, we placed six cotton threads (2–4 mm in diameter) in parallel from the top to the bottom of the phantom. Because of some technical difficulties, we did not measure the conductivity values of the cotton threads. We assume that the threads have lower conductivity than the sponges because the threads have a higher physical density compared to the sponges.

To measure the internal magnetic field components \( B_{i1} \) and \( B_{i2} \), we used an MR current density imaging (MRCDI) pulse sequence with a multislice flow-compensated spin-echo imaging technique (17,18). For the MRCDI pulse sequence, we applied a bipolar current pulse with current-driving circuitry. Since the current-driving circuit was operated in a constant-current mode, the amount of injection current was kept constant despite some irregularity in the charge exchanges between the metal plates and the electrolyte solution. The pulse width of the bipolar current was set to 48 ms in all of the experiments. With two independent MRCDI scans, we measured two internal magnetic fields \( B_{i1} \) and \( B_{i2} \) produced by currents \( I_1 \) and \( I_2 \) (Fig. 1a). During the current injection, we measured the voltage between the two electrodes, which was used in the conductivity image reconstruction.
We performed the MRCDI scans with a 3.0 Tesla MRI system (Magnum 3.0; Medinus Inc., Korea), and acquired the MRI signals using a birdcage RF coil (diameter = 30 cm). The repetition time (TR) and echo time (TE) were 1400 ms and 60 ms, respectively. The slice thickness was 5 mm with no slice gaps. We applied the oversampling technique in both the readout and phase-encoding directions to reduce truncation artifacts in phase images. Phase images (matrix size = 128 × 128) were calculated from the k-space data (matrix = 256 × 256). The total MRCDI scan time to measure the two internal magnetic fields was about 12 min. To unwrap the phase images, we used Goldstein’s branch-cut algorithm (19). We converted the phase images into magnetic flux density images by the use of appropriate scaling.

RESULTS

Figure 2 shows a typical magnitude image of the conductivity phantom and the corresponding $B_z$ image obtained with the 3.0 Tesla MRI scanner. The image was obtained with the current $I_1$ (from electrodes $E_1$–$E_3$) of 24 mA. In Fig. 2a, we can see that the recessed electrodes are quite useful for obtaining artifact-free images inside the cylindrical region of the phantom. We obtained the $B_z$ image in Fig. 2b from the corresponding phase image by using the phase unwrapping algorithm and appropriate scaling. Inside the circular region in Fig. 2b, the $B_z$ values range from $-155$ nT to 155 nT. Since we need to measure only $B_z$ in MREIT with the harmonic $B_z$ algorithm, we did not apply the object-
The harmonic $B_z$ algorithm is weak against the random noise in measured $B_z$ data, since numerical differentiations tend to amplify the noise. We are now devising different algorithms that require a single or no numerical differentiation.

Figure 3 shows bright spot artifacts in the peripheral regions between the electrodes where the two currents are almost collinear and their densities are very low. One can reduce these artifacts by using more than four electrodes evenly spaced around the object. The electrical safety guideline at the low-frequency range is specified as 100 mA/m$^2$ (20). Considering the cross-sectional area at the middle of the phantom, this guideline implies that the amount of injection current should not exceed 1.96 mA. Therefore, the noise performance should be further improved before conductivity imaging can be used in practice. For better noise performance, the SNR of $B_z$ images should be improved first. High-SNR MRI techniques, such as fast spin echo (FSE) or gradient echo (GE) imaging sequences, will be considered in future studies. In addition to high-SNR $B_z$ image acquisition, we are now developing effective denoising techniques based on underlying physical principles, such as $\nabla \cdot \mathbf{J} = 0$ and $\nabla \cdot \mathbf{B} = 0$. The current version of the harmonic $B_z$ algorithm is based on the assumption of isotropic conductivity. Since many kinds of tissues have anisotropic conductivity (21), future studies should also take into account the anisotropy of tissues.

REFERENCES