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Noninvasive Imaging of Head-Brain Conductivity Profiles Using Magnetic Resonance Electrical Impedance Imaging

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Abstract

Magnetic resonance electrical impedance tomography (MREIT) is a recently introduced non-invasive conductivity imaging modality, which combines the magnetic resonance current density imaging (CDI) and the traditional electrical impedance tomography (EIT) techniques. MREIT is aimed at providing high spatial resolution images of electrical conductivity, by avoiding solving the well-known ill-posed problem in the traditional EIT. In this paper, we review our research activities in MREIT imaging of head-brain tissue conductivity profiles. We have developed several imaging algorithms and conducted a series of computer simulations for MREIT imaging of the head and brain tissues. Our work suggests MREIT brain imaging may become a useful tool in imaging conductivity distributions of the brain and head.

Introduction

The electrical conductivity of biological tissue reflects structural, functional and pathological conditions of the tissues, and provides valuable diagnostic information [1–2]. In the past two decades, significant efforts have been made to produce cross-sectional images of conductivity distribution inside the human body by means of the electrical impedance tomography (EIT) [3–6] and its variants using magnetic induction [7]. While EIT is cost effective and provides dynamic information with regard to tissue conductivities, it is currently limited by its low spatial resolution due to the surface voltage measurements and the need to solve an ill-posed inverse problem.

Based upon the principles of magnetic resonance current density imaging (MRCDI) [8] and EIT, the magnetic resonance electrical impedance tomography (MREIT) has been pursued for the static or absolute conductivity imaging [9–11]. In comparison to the body surface voltage measurements in EIT, MREIT measures the magnetic flux density of the disturbance induced by current injection with the aid of a magnetic resonance imaging (MRI) system. One of the merits of MREIT is its high spatial resolution in obtaining conductivity images.

The past several years have witnessed considerable progress in MREIT research [12–26]. In this short paper, we review our efforts in MREIT imaging of electrical conductivity of human head and brain [15–16, 19–20, 22–26]. Our research was motivated by the need of non-invasively estimating conductivity profiles of the head tissue for solving the forward and inverse problems of electroencephalogram (EEG) and magnetoencephalogram (MEG) [1–2, 27]. We have developed several imaging algorithms, conducted a series of computer simulations to demonstrate the feasibility and assess the performance of the proposed MREIT algorithms in estimating head tissue conductivity profiles.

Principles of MREIT

Fig. 1 illustrates the MREIT experimental procedure to image the conductivity distribution within a human head. Using a constant current source and a pair of surface electrodes, a rectangular bipolar current [19] can be injected into the scalp during a spin echo pulse sequence. During MR imaging, the injection current induces a magnetic flux density $\mathbf{B}$ which causes phase changes to the main magnetic field. With the MRI measured phase shifts, the current induced magnetic flux density $\mathbf{B}$ inside the head can be computed [8]. The current density $\mathbf{J}$ can be computed from $\mathbf{B}$ based on Ampere's Law $\mathbf{J} = \nabla \times \mathbf{B} / \mu_0$. A technical challenge is that, using $\mathbf{J} = \nabla \times \mathbf{B} / \mu_0$, all the three components of $\mathbf{B} = (B_x, B_y, B_z)$ need to be measured. As the MRI scanner measures only one component of $\mathbf{B}$ at a time, which is parallel to the direction of the main magnetic field B_0 of the scanner, it is impractical to rotate a human subject inside the MRI scanner bore. Therefore, much of the recent efforts has focused on developing MREIT algorithms which can reconstruct the conductivity image from only one component of $\mathbf{B}$ (usually referred as B_z) that is parallel to B_0 . With current injection, the information embedded in the measured B_z can be analyzed. Mathematical techniques have been employed to construct the relations between B_z and the conductivity distribution, such as the harmonic B_z algorithm [12], the algebraic reconstruction algorithm [13], and the current density reconstruction algorithm [25].

The MREIT forward problem is defined as the computation of the magnetic flux density distribution with a known conductivity distribution σ and boundary conditions. In ω , a bounded and electrically conductive domain in $\mathbb{R}^3$ with boundary Γ , the electrical potential distribution ϕ obeys Laplace's equation and Neumann boundary conditions as follows:

$$\begin{cases} \nabla \cdot (\sigma \nabla \Phi) = 0 & \text{in } \Omega \\ -\sigma \nabla \Phi \cdot \mathbf{n} = \mathbf{J}_{\text{inj}} & \text{on } \Gamma \end{cases} \quad (1)$$

where $\mathbf{n}$ is the unit outward normal vector and $\mathbf{J}_{\text{inj}}$ the injected body surface current density which is non-zero only on the current injection electrodes. Then the current density distribution can be derived from the potential distribution as:

$$\mathbf{J} = -\sigma \nabla \Phi \quad (2)$$

The magnetic flux density $\mathbf{B}$, due to $\mathbf{J}$, can be obtained according to the Biot-Savart Law

$$\mathbf{B}(\mathbf{r}) = \frac{\mu_0}{4\pi} \int_{\Omega} \mathbf{J}(\mathbf{r}') \times \frac{\mathbf{r} - \mathbf{r}'}{|\mathbf{r} - \mathbf{r}'|^3} dV' \quad (3)$$

where $\mathbf{r}$ and $\mathbf{r}'$ refer to the field and source points located in ω , and μ_0 the magnetic permeability of vacuum.

Given the information of measured magnetic flux density and injected current, the MREIT inverse problem aims to estimate the unknown conductivity values of the interested tissues. There are several algorithms to solve the inverse problem of MREIT, either by solving the explicit relationship between the magnetic flux density and conductivity distribution [12–13, 25], or by establishing a mapping relationship between them [15,19].

Reconstruction of Head Conductivity Profiles by MREIT

We have developed several algorithms for MREIT imaging of head and brain conductivity profiles, and conducted computer simulation studies to evaluate the performance of the developed algorithms. We used the finite element method (FEM) to solve the MREIT forward problem. Two widely used head models were considered: a 3-layer spherical head model and a 3-layer realistic-geometry (RG) head model, representing significant conductivity profiles consisting of the brain, scalp and skull. Fig. 2 shows the finite element models of these two head models. In our computer simulations, we injected current of amplitude of 4–5 mA since 5 mA is considered to be the upper safe limit for human beings (IEC Criteria 1973). The ANSYS 10.0 (ANSYS Inc., PA, USA) software was used to simulate the current induced B_z , which was used as “simulated” B_z .

Estimation of conductivity in a piece-wise homogeneous head model

Assuming each tissue compartment of the head models to be homogeneous and the conductivity to be uniformly distributed [1–2], the problem becomes to estimate the conductivity values $\sigma = [\sigma_{brain}, \sigma_{skull}, \sigma_{scalp}]$ within the head. We normalized all the conductivity values with reference to σ_{brain} for the sake of simplicity [15], and the target conductivity value σ was set to be [1, 1/25, 1] or [1, 1/15, 1].

Given the target value of σ^* , the “MRI measured” B_z^* was obtained by solving the MREIT forward problem. Then a system was constructed and optimized in the MREIT reconstruction problem as follows: (1) As the system input parameter σ , an initial guess of the conductivity value was made to start the inverse procedure. (2) By solving the forward problem, the corresponding magnetic flux density B_z was derived, and an objective function $f(\sigma)$, as the system output, was calculated to assess the dissimilarity between B_z^* and B_z . (3) With the input-output pairs σ and $f(\sigma)$, we constructed the system representing the output $f(\sigma)$ as a function of input conductivity distribution σ (Fig. 3). (4) The conductivity value was estimated when the objective function $f(\sigma)$ was minimized [15,19]. The objective function $f(\sigma)$ was defined as

$$f(\sigma) = (1 - CC(B_z^*, B_z))^2 + RE(B_z^*, B_z) \quad (4)$$

where $CC(B_z^*, B_z)$ and $RE(B_z^*, B_z)$ were the correlation coefficient (CC) and relative error (RE) between B_z^* and B_z , respectively.

We have developed three algorithms to optimize the above objective function (Eq. (4)): the radial basis function (RBF) algorithm which is an artificial neural network using a set of basis functions in the hidden units [15]; the response surface methodology (RSM) algorithm which is a classic method of curve fitting based on the approximation of the objective function by a low-order polynomial [19]; and the adaptive neuro-fuzzy inference system (ANFIS) algorithm which is a multilayer neural network-based fuzzy system [23]. The simplex method was used to search for the optimal conductivity value. Gaussian white noise (GWN) was added with various noise levels to simulate noise-contaminated B_z^* measurements. The standard deviation of noise s_B was given by $s_B = 1/2\gamma T_c SNR$, where γ is the gyromagnetic ratio of hydrogen, T is the duration of the injection current pulse, and SNR is the signal-to-noise ratio of the MR magnitude image which was set to be 80, 60, 40, 20 and 15, respectively. In order to assess the clinical applicability, GWNs with standard deviation of 5mm and 10mm [15–16] were also added to the electrode positions to simulate the effects of electrode position uncertainty.

Two numerical simulations were conducted on the two head models: single-variable and three-variable simulations. In the single-variable simulation, only σ_{skull} was assumed to be unknown, while in the three-variable simulations all the three conductivity components $\sigma = [\sigma_{brain}, \sigma_{skull}, \sigma_{scalp}]$ were unknown. In these simulations, a bipolar rectangular current of 4 mA was applied to the head model through a pair of opposite electrodes as shown in Fig. 2(b) and (d). In the single-variable simulation on the RG head model, when the target value of σ_{skull} was set to 1/15 and 1/25 (in reference to brain conductivity), the estimation errors (RE) of ANFIS-MREIT, RBF-MREIT and RSM-MREIT algorithms are shown in Fig. 4. Note that all REs are less than 10% under five SNR levels of MR magnitude image, but the ANFIS-MREIT returned much smaller errors compared with two other algorithms. Applying the RBF-MREIT and RSM-MREIT algorithms on the RG head model, the estimation errors (RE) of the three-variable simulation were less than 17% and 12%, respectively, suggesting the applicability of these MRIET algorithms to the estimation of important conductivity values which were used in the EEG forward/inverse problem.

Estimation of conductivity in an inhomogeneous head model

In a human head, the tissue conductivity distribution is not uniform [1–2,26]. The problem can be considered to estimate the conductivity distribution matrix

$\sigma = \{\sigma_{tissue}^i\} = \{\sigma_{brain}^o, [\sigma_{skull}^p], [\sigma_{scalp}^q]\}$, in which o, p, q denote the meshed element number for each tissue compartment, and σ_{tissue}^i is the conductivity value of the i -th element in the entire head model. In this simulation, we used the 3-layer spherical head model (Fig. 2). σ_{tissue}^i was also normalized as a ratio in reference to the target brain conductivity. The distributions of $\sigma_{brain}^o, \sigma_{skull}^p$ and σ_{scalp}^q were simulated by varying within ranges enclosing the target values of 1, 1/15 and 1, respectively. Each conductivity value was set randomly within the following ranges: $\sigma_{brain}^o \in (0.9, 1.1)$, $\sigma_{skull}^p \in (1/14.9, 1/15.1)$ and $\sigma_{scalp}^q \in (0.9, 1.1)$, to test the robust of the MREIT algorithms to image the inhomogeneous conductivity distribution.

We used the B_y - B_z based algebraic reconstruction algorithm [20] in this simulation. With one rotation in the MRI scanner, we could measure B_y and B_z components; and by solving the MREIT forward problem using the FEM, we could compute the current density $\mathbf{J}$ corresponding to a conductivity distribution σ . Equation (5) suggests the relationship among σ , $\mathbf{J}$ and $\mathbf{B}$, where S is defined by $1/\sigma = e^S$ [13]. Unwrapping the circled rows, a non-linear matrix equation was composed with B_y, B_z and $\mathbf{J}$, then we could derive a new conductivity distribution σ_{new} by solving S with the known B_y, B_z and $\mathbf{J}$. We used this σ_{new} in the forward problem again and solved the forward problem iteratively. The i th solution σ_i , when the difference between σ_i and σ_{i-1} was lower than a tolerable error, was considered as the reconstructed conductivity distribution.

$$\begin{bmatrix} 0 & J_z & -J_y \\ -J_z & 0 & J_x \\ J_y & -J_x & 0 \end{bmatrix} \begin{bmatrix} \frac{\partial S}{\partial x} \\ \frac{\partial S}{\partial y} \\ \frac{\partial S}{\partial z} \end{bmatrix} = - \begin{bmatrix} \frac{\partial J_z}{\partial y} - \frac{\partial J_y}{\partial z} \\ \frac{\partial J_x}{\partial z} - \frac{\partial J_z}{\partial x} \\ \frac{\partial J_y}{\partial x} - \frac{\partial J_x}{\partial y} \end{bmatrix} = \frac{1}{\mu_0} \begin{bmatrix} \nabla^2 B_x \\ \nabla^2 B_y \\ \nabla^2 B_z \end{bmatrix} \rightarrow \begin{cases} \frac{\partial S}{\partial z} J_x - \frac{\partial S}{\partial x} J_z = \frac{\nabla^2 B_y}{\mu_0} \\ \frac{\partial S}{\partial x} J_y - \frac{\partial S}{\partial y} J_x = \frac{\nabla^2 B_z}{\mu_0} \end{cases}$$

(5)

Fig. 5 depicts the cross-sectional conductivity images of the 3-layer spherical head model with GWN at three SNRs [20]. When SNR is 10, the RE of estimated conductivity was 7.67% for the brain region and 7.84% for the scalp region, which indicates the B_y - B_z based algebraic algorithm can obtain reasonable results for inhomogeneous conductivity distributions, when variation in conductivity is close to the target value of the conductivity of each compartment.

Detecting of brain tissue conductivity associated with pathological conditions

It is well known that, under some physiological phenomena, brain activity will cause local and temporal conductivity changes. For example, cell swelling reduces the extracellular space and increases the bulk tissue impedance as much as 50%, therefore the ischemic tissue can be characterized by abnormally high impedance [16]. We have examined the feasibility of using MREIT to detect conductivity changes associated with brain pathological conditions, and the two-step method [22] was utilized: the inhomogeneous conductivity distribution σ of the encephalic pathological tissue, which was assumed to be homogeneous, was first estimated by the RBF-MREIT algorithm; then the genetic algorithm (GA) was applied to estimate σ_{tissue}^i for each element within each tissue of the FEM head model [24]. The RG head model was utilized and the pathological tissue was indicated by an anomaly as shown in Fig. 6(a). Figs. 6(b) – (f) show the cross-sectional images of reconstructed anomaly with GWN added at five SNRs. Fig. 6 suggests the feasibility of detecting anomaly using MREIT.

Discussion

MREIT promises to provide high spatial resolution in imaging electrical conductivity of a biological system with the aid of MRI. By avoiding solving an ill-posed inverse problem in traditional electrical impedance tomography (EIT), MREIT has promising features among impedance imaging approaches. In this short paper, we have reviewed the computational efforts, which our group has made in the past several years, in imaging and estimating conductivity profiles of the head. As our simulation results indicate, the B_z -based MREIT algorithms furnish us with feasible and practical approaches to reconstruct the conductivity distribution within a head. Note also, that when the relative conductivity is concerned, the algorithms we have developed do not require voltage measurement using surface electrodes.

Since a large variation exists among reported data [27], accurate estimation of the brain-to-skull conductivity ratio is important to improve the accuracy of brain source localization. While recent experimental studies suggest that the brain-to-skull conductivity ratio be 20 from simultaneous intra- and extra-cranial recordings in human subjects [29], it is desirable that such information can be obtained noninvasively for each individual. In the series of computer simulation studies we have conducted, it is clear that MREIT provides useful information with regard to the important conductivity values of head compartments which are widely used in EEG/MEG source localization and imaging. It is noteworthy that application of conventional EIT to head conductivity imaging is challenged by the low skull conductivity because much of the current is shunted through the scalp without entering into the brain. MREIT has the unique feature to measure magnetic flux density throughout the 3-dimensional space without severe influence by the low conductivity skull. Our simulation results support this notion.

While our MREIT head-brain imaging was motivated by the application to EEG/MEG forward/inverse problems, the application of MREIT to the brain research is not limited to the source localization problem. High resolution conductivity images of the brain and head will play an important role in neuroscience research and may have important applications in clinical neurology and neurophysiology.

The major limitation of MREIT is the need of current injection with sufficient large amplitude, in order to achieve good signal-to-noise ratio. This is also the reason that there are currently

no in vivo clinical studies reported in human subjects. Several studies were carried out using phantoms to evaluate the performance of MREIT, and the peak value of 5–25 mA was typically adopted as the current amplitude. Recently an *in-vivo* tumor bearing experiment on rats was conducted for breast tumor localization with a current injection of amplitude of 1 mA [18]. In our simulation studies, a current injection of 4 mA or 5 mA was applied, with 5 mA being thought the upper safe limit for human beings (IEC Criteria 1973). Further efforts should be made to substantially reduce the amount of current injection in order to move MREIT into clinical applications.

Due to the present limitation of high level current injection, we conducted a series of computer simulations on MREIT brain imaging research. We used ANSYS software to simulate current injection configurations and B_z measurement, and conducted MRI image-processing procedures in a “virtual measurement” scheme. However, the feasibility of MREIT brain imaging in an experimental setup has been recently suggested in which high level of current injection was used [21]. Future investigations should be directed at experimentally demonstrating MREIT brain imaging with low current injection for potential clinical applications.

In summary, we have pursued, since 2004, the electrical impedance imaging of head-brain tissue conductivity profiles by means of the MREIT. We have developed several MREIT algorithms for head-brain imaging and demonstrated the feasibility and merits of imaging head-brain conductivity profiles by means of MREIT in a computational setting. While further development is needed in order to apply MREIT in a clinical setting, experimental evaluation is needed to fully assess the utility and applicability of the developed MREIT algorithms. Our work suggests that MREIT brain imaging merit further investigation and may become a useful tool in imaging conductivity distributions of the brain and head.

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Figure 1.

Figure 2.

Figure 3.

Figure 4.

Figure 5.

Figure 6.