Induced Current Magnetic Resonance Electrical Impedance Tomography with z-Gradient Coil

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Abstract— **Magnetic Resonance Electrical Impedance Tomography (MREIT) is a medical imaging method that provides images of electrical conductivity at low frequencies (0-1 kHz). In MREIT, electrical current is applied to the body via surface electrodes and corresponding magnetic flux density is measured by means of Magnetic Resonance (MR) phase imaging techniques. By utilizing the magnetic flux density measurements and surface potential measurements images of true conductivity distribution can be reconstructed. In order to overcome difficulties regarding current application via surface electrodes, Induced Current MREIT (ICMREIT) have been proposed in the past. In ICMREIT, electrical currents and corresponding magnetic flux density are generated in the object through electromagnetic induction by means of externally placed coils driven with time varying currents. In this study, use of z-gradient, z-Helmholtz, and circular coil configurations in ICMREIT are proposed and investigated. Finite Element Method (FEM) is used to solve the forward problem of ICMREIT. Consequently, excitation performances and clinical applicability of different coil configurations are analyzed.**

I. Introduction

Electrical conductivity properties of biological tissues vary between the tissues and according to their physiological activity. Therefore, measuring electrical conductivity of biological tissues may provide valuable diagnostic information. Magnetic Resonance Electrical Impedance Tomography (MREIT) is a medical imaging modality that utilizes Magnetic Resonance (MR) phase imaging techniques and surface potential measurements to reconstruct images of true conductivity distribution within objects that contain MR active nuclei. In this method, low frequency (0-1kHz) electrical current is applied to the body via surface electrodes in synchrony with a spin-echo pulse sequence. Magnetic field created by the applied current acts like a gradient field and induces a phase in the MR signal. Therefore, the phase term related to the current induced magnetic field can be reconstructed. Current density distribution which, is a function of conductivity distribution within the object, can be obtained from the phase image. By using the current density information and measured surface potentials, conductivity images could be obtained. Necessity of applying electrical current by means of electrodes introduces a major drawback in application of MREIT.

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In addition to that, this method has also clinical drawbacks due to high current densities in the regions that are close to the electrodes. Further, it is difficult to image regions surrounded by highly resistive tissues such as brain which is surrounded by skull [1].

In order to overcome these difficulties, Özparlak and Ider proposed Induced Current Magnetic Resonance Electrical Impedance Tomography (ICMREIT), in a simulation study [2]. In ICMREIT, a coil is placed in the vicinity of the body to be imaged and it is excited by time varying electrical current in synchrony with an appropriate MRI pulse sequence, which induces an eddy current distribution within the body. Similar to MREIT, the eddy current in the body creates a secondary magnetic field and induces a phase term in the MR signal, which could be measured by MREIT phase imaging techniques. By utilizing the MR phase images conductivity images could be reconstructed. One of the potential difficulties of ICMREIT is the presence of additional coils within a conventional MR Imaging (MRI) hardware. From a practical point of view, the ICMREIT coils should be located in the outer domain of the RF coil in order to acquire MRI signal properly. On the other hand, driving the ICMREIT coil at high Ampere-turns also seems to be another difficulty due to the need for a high capacity and MR compatible AC current supply. Using the gradient coils that are readily available in an MRI scanner could be a potential solution [3-4].

In this study, forward problem of ICMREIT is introduced briefly and a phase image calculation method is explained. Afterwards, the forward problem is solved for a z-gradient, z-Helmholtz, and circular coil configurations by using Finite Element Method (FEM). Consequently, simulation results for different coil configurations are compared with each other and these coil configurations are evaluated with respect to clinical applicability.

II. Forward Problem

The forward problem of ICMREIT can be defined as generating a secondary magnetic flux density (B_{sz}) distribution through a volume conductor by means of exciting a coil configuration with time varying current. When a coil configuration around a cylindrical conductor as shown in Fig. 1 is excited with time varying current, a primary magnetic flux density (B_{pz}) is created throughout the volume conductor. Since B_{pz} has also time varying nature, an eddy current distribution (J_{eddy}) is created in the conductor by means of electromagnetic induction. The induced J_{eddy} results in B_{sz} distribution through the object.

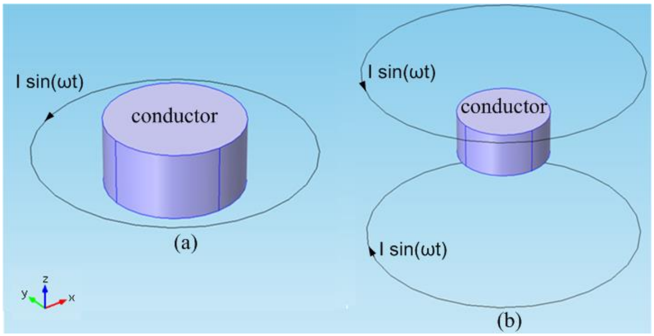

Figure 1. Cylindirical conductor surrounded with circular (a) and z-gradient coils.

Mathematically, the forward problem of ICMREIT can be explained starting from Ampere's Law as shown in (1) in which H is the magnetic field intensity, σ and ε are the conductivity and permittivity of the media, respectively, ω is the electrical frequency, E is the electric field, and J_e is the external current density.

$$\nabla \times \overline{H} = (\sigma + j\omega\varepsilon)\overline{E} + \overline{J}_e \quad (1).$$

If E is expressed as a function of vector magnetic potential (A) as shown in (2) and the relationship between H and B is used as shown in (3), the Ampere's Law expression can be modified as shown in (4). In (3-4), μ represents the permeability of the media.

$$\overline{E} = -j\omega\overline{A} \quad (2)$$

$$\overline{H} = \mu^{-1}\overline{B} \quad (3)$$

$$\left(j\omega\sigma - \omega^2\varepsilon\right)\overline{A} + \nabla \times \left(\mu^{-1}\overline{B}\right) = \overline{J}_e \quad (4)$$

B can be expressed as the curl of A as shown in (5) and curl of B is expressed in (6). By utilizing (5-6) with the Coulomb's gauge expressed in (7), the final differential equation representing the forward problem can be obtained as shown in (8).

$$\overline{B} = \nabla \times \overline{A} \quad (5)$$

$$\left(\nabla \times \nabla \times \overline{A}\right) = \nabla\left(\nabla \bullet \overline{A}\right) - \nabla^2\overline{A} \quad (6)$$

$$\nabla \bullet \overline{A} = 0 \quad (7)$$

$$\left(j\omega\sigma - \omega^2\varepsilon\right)\overline{A} - \mu^{-1}\nabla^2\overline{A} = \overline{J}_e \quad (8)$$

In (8), σ, ε, ω, and J_e are the known parameters, whereas A is the unknown. Therefore, (8) is solved for A by using FEM and afterwards, B calculation is performed by utilizing (5). For the outer surfaces of the solution domain, magnetic insulation type boundary condition can be applied as shown in (9) and zero initial condition can be used for A as shown in (10).

$$\overline{n} \times \overline{A} = 0 \quad (9)$$

$$\overline{A} = \begin{bmatrix} 0 & 0 & 0 \end{bmatrix}^T \quad (10)$$

The forward problem shown in (8) together with the boundary conditions shown in (9-10) is defined in a FEM model. In the FEM model, coil structures are positioned such that the majority of the magnetic flux density is focused in z direction, in parallel with the main magnetic field of the MRI systems. The solutions are obtained for full magnetic flux density distribution. In order to obtain the secondary magnetic flux density (B_{sz}), two simulations are performed. In the first simulation, B field is obtained for the case where the conductivity of the object is made equal to the real case. In the second simulation, B field is obtained for the case where the conductivity of the object is made equal to the media such as air. Consequently, the difference of the two B fields is calculated in order to obtain B_{sz} distribution.

III. MRI Pulse Sequence and Phase Calculation

In ICMREIT, the measured quantity is the phase accumulated during the MRI experiment. Özparlak and İder [2] proposed a spin-echo based MRI pulse sequence for ICMREIT as shown in Fig. 2. In general the phase accumulated in the MR image due to a generated B_z field can be expressed as shown in (11).

$$\varphi = \gamma \; B_{z\_av} \; t_{app} \quad (11)$$

In (11), φ is the accumulated phase, γ is the gyromagnetic ratio, B_{z_av} is the average B_z field, and t_{app} is the interval of coil excitation. The contribution of B_{pz} to B_{z_av} is zero since it is in phase with the excitation current ($i_{excitation}$) and 180^o RF pulses are applied at each peak of B_{pz} waveform. On the other hand, the contribution of B_{sz} to B_{z_av} is non-zero since it is generated by electromagnetic induction and out of phase with B_{pz} by 90^o. By replacing the B_{z_av} value with the average of B_{sz} and considering the number of 180^o RF pulses (N), the accumulated phase can be expressed as shown in (12) [2].

$$\varphi = (-1)^N \gamma \frac{2B_{sz\_peak}}{\pi} t_{app} \quad (12)$$

If the pulse sequence is applied twice in which $i_{excitation}$ is reversed in the second application and the difference operation between the two MR phase images is performed, the obtained phase could be doubled as shown in (13) [2].

$$\varphi = 2 \; (-1)^N \gamma \frac{2B_{sz\_peak}}{\pi} t_{app} \quad (13)$$

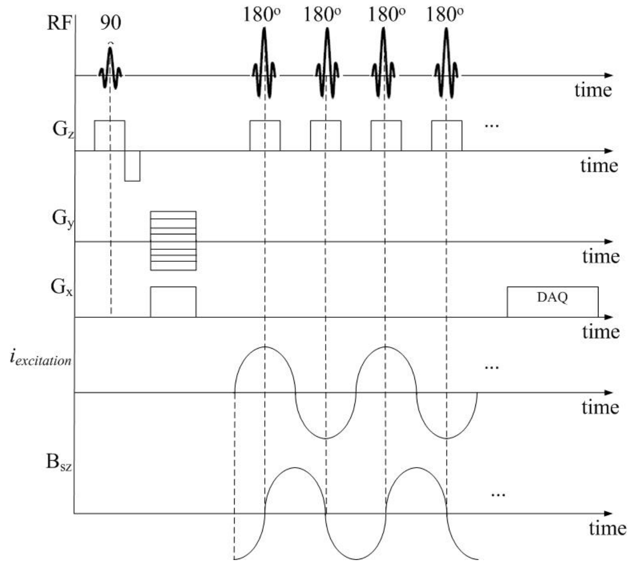

Figure 2. ICMREIT pulse sequence proposed by Özparlak and İder [2]

IV. SIMULATION RESULTS

As mentioned in the previous sections, circular, z-gradient, and z-Helmholtz coils are modeled in FEM. Circular and z-gradient coils with a cylindrical conductor are shown in Fig. 1. The simulation parameters are summarized in Table I-II. As shown in Table I, two coil currents are applied in opposite direction for the z-gradient coil in order to obtain a gradient field, whereas coil currents are applied in the same direction for the z-Helmholtz coil. With the input Ampere-turns value of the z-gradient coil, a gradient strength of 45 mT/m is obtained. B_{sz} distributions are calculated for the central slice for the circular coil, whereas for the z-gradient and z-Helmholtz coils, it is calculated for the uppermost slice. After the B_{sz} distributions are obtained, corresponding phase images are calculated using (13). In the phase image calculation procedure t_{app} and N are chosen as 20 ms and 200, respectively. Simulation results are shown in Fig. 3 and summarized in Table III.

TABLE I. COIL PARAMETERS

Simulation Model	Location at z-axis (cm)	Diameter (cm)	Input Ampere-Turns	Input Frequency (kHz)
Circular	0	40	1000	1
z-Gradient	±20		± 12700	
z-Helmholtz			12700	

TABLE II. CONDUCTOR PARAMETERS

Simulation Model	Size [r x h] (cm)	Location at z-axis (cm)	σ (S/m)	Selected xy slice at z-axis (cm)	Area of the selected xy slice (cm^2)
Circular	10 x 10	±5	1	0	100
z Gradient		0-10		10	
z Helmholtz					

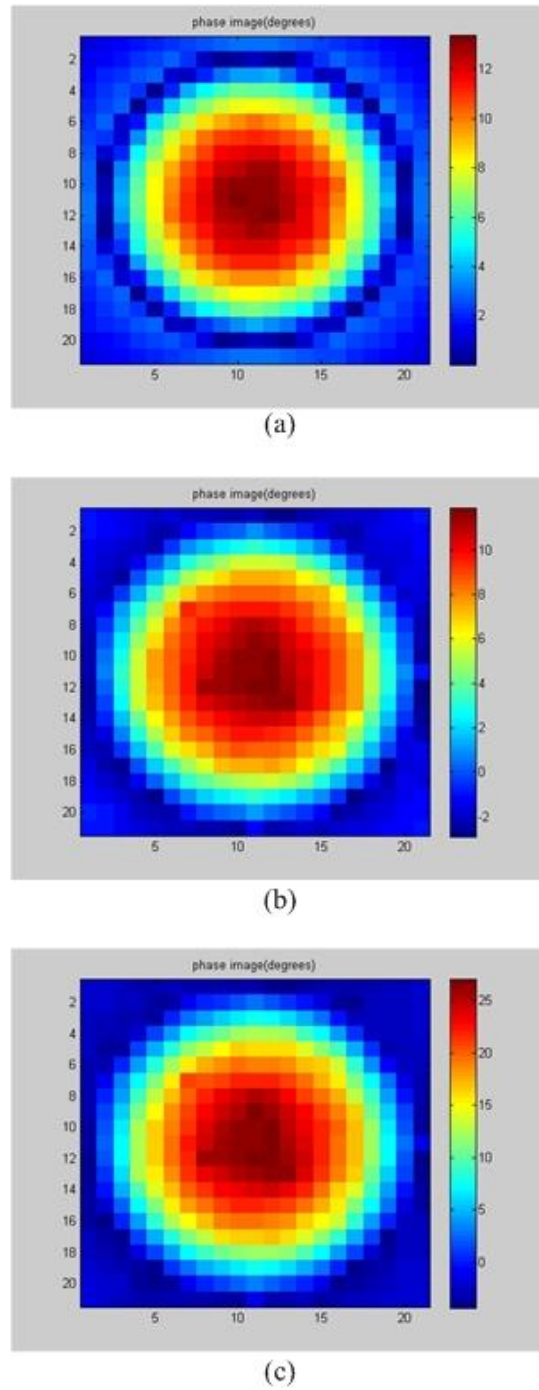

Figure 3. Phase images for circular (a), z-gradient (b), and z-Helmholtz (c) coils.

TABLE III. SIMULATION RESULTS

	Absolute Value of Phase (degrees)			E (V/m)
	Max.	Min.	Mean	Max.
Circular	13.36	3.02	5.69	0.9744
z Gradient	11.92	2.90	5.09	1.7869
z Helmholtz	27.06	4.37	11.84	2.85

V. Discussion

As seen in the simulation results given in Fig. 3 induced phase is greater at the center of the conductor. This is a reasonable observation as shown in Fig. 4 where eddy currents flow azimuthally for a B_{pz} distribution. When B_{sz} values at points A and B are compared, it is seen that all eddy current patterns contributes to the magnetic field on A in the same direction, whereas the contribution of current patterns J_1 and J_2-J_3 to the point B are opposite. Therefore, B_{sz} value at A is greater than the B_{sz} value at point B. The simulation results also show that the phase distribution has homogenous charateristics in the center of the object.

Induced phase values calculated for different coil geometries are summarized in Table III. The maximum, the minimum, and the mean phase values vary between 11.9-27, 2.9-4.4, and 5.0-11.8 degrees, respectively. The accumulated phase values obtained from the simulations seem to be measurable for all configurations which shows the feasibility of ICMREIT. The obtained phase values could be increased linearly by increasing t_{app} parameter of (13). When the average phase obtained from the three simulation models are compared with each other, it is seen that z-Helmholtz coil generates the greatest phase. The phase generated by the z-gradient coil is at least 2 times lower than the phase generated by the z-Helmholtz coil. Therefore, if it is possible to drive the coils of z-gradient system of an MRI scanner in the same direction, greater ICMREIT sensitivity could be obtained. The main advantage of using z-gradient coil of an MRI system is the utilization of readily available components.

It is also observed that the phase generated by the circular coil is greater than the phase generated by the z-gradient coil, although the input Ampere-turns of the circular coil is 12.7 times lower than the input Ampere-turns value of the z-gradient coil. Circular coil generates phase of the same order as the phase generated by the z-gradient and z-Helmholtz coils, since it is closer to the object. Therefore, it could be concluded that by using circular coils greater ICMREIT sensitivity could be obtained. In other words, by using circular coils, measurable phase could be generated with less input Ampere-turns. Since the input Ampere-turns requirement will decrease for the circular coil, input frequency could also be increased which will further increase the sensitivity of ICMREIT. Although the use of circular coil in ICMREIT seems to be advantageous in terms of sensitivity, the practical implementation of circular coil does not seem straight-forward. A high capacity, MR compatible AC current supply is necessarry in order to conduct ICMREIT experiments with circular coil. In addition, the circular coil should be placed in the outer domain of the RF coil of the MRI scanner. Otherwise, imaging artifacts will be present. Therefore, it could be concluded that RF and circular ICMREIT coils should be designed together if circular coils are adopted to be used in ICMREIT.

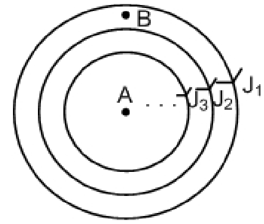

Figure 4. Eddy currents in the cross section of a cylindirical conductor.

In Table III, the maximum electric field values for the three coil geometries are also presented. Borsic and Paulsen stated that the maximum electric field value inside the body should be below 3.75 V/m for ICMREIT experiments at 1 kHz according to IEC60601-2-23 standard [5]. The simulation results show that the maximum induced electric field values in the conductor object are below the maximum induced electric field limitation of the IEC60601-2-23 standard for clinical use.

VI. Conclusion

In this study, forward problem of ICMREIT is presented and solved using FEM for circular, z-gradient, and z-Helmholtz coils with a cylindrical conductor object. Calculated phase distributions show that, all three coils seem to generate measurable phase distributions. It is clear that the use of z-gradient coil seems to be advantageous in terms of practical implementation, since there is no need for an additional AC current source and coil. It is also seen that if it is possible to drive the coils of z-gradient system of an MRI scanner in the same direction, more ICMREIT sensitivity could be obtained. When compared with the z-gradient and z-Helmholtz coils, circular coil has potential to generate higher sensitivity with reduced input Ampere-turns. However, for this configuration, state of the art AC current supply and coil design is necessary. Finally, based on the performed simulation study it can be concluded that induced electric field levels of ICMREIT seems to be within the safety limits.

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