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Magnetic Flux Density Measurement with Balanced Steady State Free Precession Pulse Sequence for MREIT: a Simulation Study

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***Abstract*—Magnetic Resonance Electrical Impedance Tomography (MREIT) utilizes the magnetic flux density B_z, generated due to current injection, to find conductivity distribution inside an object. This B_z can be measured from MR phase images using spin echo pulse sequence. The SNR of B_z and the sensitivity of phase produced by B_z in MR phase image are critical in deciding the resolution of MREIT conductivity images. The conventional spin echo based data acquisition has poor phase sensitivity to current injection. Longer scan time is needed to acquire data with higher SNR. We propose a balanced steady state free precession (b-SSFP) based pulse sequence which is highly sensitive to small off-resonance phase changes. A procedure to reconstruct B_z from MR signal obtained with b-SSFP sequence is described. Phases for b-SSFP signals for two conductivity phantoms of TX 151 and Gelatin are simulated from the mathematical models of b-SSFP signal. It was observed that the phase changes obtained from b-SSFP pulse sequence are highly sensitive to current injection and hence would produce higher magnetic flux density. However, the b-SSFP signal is dependent on magnetic field inhomogeneity and the signal deteriorated highly for small offset from resonance frequency. The simulation results show that the b-SSFP sequence can be utilized for conductivity imaging of a local region where magnetic field inhomogeneity is small. A proper shimming of magnet is recommended before using the b-SSFP sequence.**

I. Introduction

Magnetic flux density, generated due to current injection, is utilized in Magnetic Resonance Electrical Impedance Tomography (MREIT) to reconstruct conductivity distribution inside an electrically conducting object [1-5]. For a current injection along x-y plane, the magnetic flux density, referred to as B_z, is produced along the main magnetic field in z-direction. This B_z makes extra phase in magnetic resonance images. Since MREIT image quality is strongly dependent on B_z, it is essential to measure it with high SNR. For safety purposes, injection current level is limited to 1mA in human studies. It is therefore desired to increase the sensitivity of MREIT to the injection current.

In conventional MREIT, the B_z generated by the injection current is sensitized by the transverse magnetization formed by 90^0 RF pulse. The phase of transverse magnetization at the data acquisition period is determined by the imaging gradients, main field inhomogeneity, and the magnetic field B_z generated by the injection current. If the spin echo pulse sequence is used for MREIT, considering two orthogonal current injections, all phases except the one from the injection currents, are cancelled out. Therefore, the phase due to the injection current can be calculated from the magnetic resonance images [4]. When transverse magnetization is used to sensitize the magnetic field, the phase is determined by $\phi = \gamma B_z T_c$ where γ is the gyromagnetic ratio of proton and T_c the current injection time [6]. With the current injection level of 1mA, the magnetic field strength of the order of 10nT is formed in an imaging region with the size of $20cm^2$. Considering the T2 decay and many other artifacts coming from long echo time, practical current injection time is limited to less than 30ms. With the maximum allowable current injection time of 30ms, the maximum phase induced by the injection current of 1mA is only of the order of $\pm\pi/39$ radians. In order to detect this small phase with substantial signal to noise ratio (SNR), higher averaging cycles are needed while acquiring MR data, which increases the scan time.

Steady state free precession (SSFP) pulse sequences are used for fast MRI [6-9]. Unlike spin echo pulse sequence where all the phase of transverse magnetization is nullified before the next 90 degree RF pulse, in SSFP sequence, the phase of the transverse magnetization is preserved in next pulsing periods thereby forming a steady state of both transverse and longitudinal magnetizations. Fast MRI using the SSFP, where all the phase is nullified before the next 90^0 RF pulse, has much higher sensitivity than other fast MRI techniques [10]. There are two kinds of SSFP pulse sequences, unbalanced and balanced SSFP (b-SSFP). Unlike unbalanced, in balanced SSFP the gradient-field-induced phase of all the spins is set to zero at the end of pulsing periods as shown in figure 1. For fast MRI, unbalanced SSFP is preferred since it is more robust to field inhomogeneity effects. In b-SSFP, field inhomogeneity gives significant effects on the SSFP signals, in amplitude and phase, since the field-inhomogeneity-induced phase is not nullified at the end of pulsing period. However, the field-inhomogeneity-induced phase has been successfully exploited in functional MRI (fMRI) in which local field inhomogeneity caused by oxygen-carrying arterial blood is used to make functional contrast. It has been reported that SSFP fMRI has great

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potentials for human brain function studies if the imaging field of view (FOV) is confined to a small regional area [8].

In this work, we propose the b-SSFP pulse sequence for MREIT to improve the phase sensitivity to current injection. Since the phase induced by the injection current at a given pulsing period is preserved at next pulsing periods, SSFP would give much higher sensitivity than conventional spin echo based MREIT. Theoretical background behind the B_z extraction from SSFP phase images is described here followed by some simulations comparing the phases obtained from two uniform conductivity phantoms of TX-151 and Gelatin.

II. METHODS

A. Theory

The transverse magnetization right after the RF pulse in b-SSFP is given by [11],

$$M_y^+(\theta) = M_o(1-E_1)\sin\alpha(1-E_2\cos\theta)/D$$

$$M_x^+(\theta) = M_o(1-E_1)\sin\alpha E_2\sin\theta/D$$

$$D = (1-E_1\cos\alpha)(1-E_2\cos\theta) - E_2(E_1-\cos\alpha)(E_2-\cos\theta)$$

$$E_1 = \exp(-TR/T_1)$$

$$E_2 = \exp(-TR/T_2)$$

where, M_o is the equilibrium magnetization, M_x^+ and M_y^+ the x and y-directional transverse magnetizations, α the flip angle, TR the repetition time, and θ the precession angle, or the total phase at the end of pulsing period of the spin. In MREIT, the phase information is used to estimate B_z. The phase of the b-SSFP signal right after the RF pulse can be given by:-

$$\phi^+(\theta) = \tan^{-1}(M_x^+/M_y^+) = \tan^{-1}\left(\frac{E_2\sin\theta}{1-E_2\cos\theta}\right)$$

In SSFP, both the magnitude and the phase of the spin are governed by the precession angle θ. For MREIT, the precession angle θ in a pulsing period TR is given by [11],

$$\theta = \int_0^{TR}\gamma B_z dt + \gamma\cdot\Delta B\cdot TR$$

where, B_z is the magnetic field induced by the injection current, and ΔB is the total field inhomogeneity arising from main field inhomogeneity, susceptibility effect and chemical shift. If we apply constant current for time T_c in a pulsing period, then

$$\theta = \gamma\cdot B_z\cdot T_c + \gamma\cdot\Delta B\cdot TR$$

If we apply the electric current constantly in a whole pulsing period and position the data acquisition window at the center of pulsing period as shown in figure 1, the phase of the MRI signal in a given pixel is given by,

$$\phi(\theta) = \tan^{-1}\left(\frac{E_2\sin\theta}{1-E_2\cos\theta}\right) + \frac{1}{2}\theta$$

It is notable that the phase ϕ at a given pixel is a nonlinear function of the precession angle θ, the T_2-decay during a repetition time and TR. Rearranging the above equation,

$$\theta_{1,2} = 2\times\tan^{-1}\left(\frac{1-E_2}{1+E_2}\tan\phi_{1,2}\right)$$

where, $\theta_{1,2}$ are the precession angles in a pulsing period in the two cases of with and without injection current, respectively, and $\phi_{1,2}$ are the corresponding phases of the b-SSFP signal. $\phi_{1,2}$ are known from the experiment. If the T_2 value is known at a given pixel, the phase change caused by the injection current θ_{Bz}, can be calculated from the previous equation as follows:-

$$\theta_{B_z} = \theta_1 - \theta_2$$

Hence, the magnetic flux density can be calculated as:-

$$B_z = \frac{\theta_{B_z}}{\gamma\cdot T_c}$$

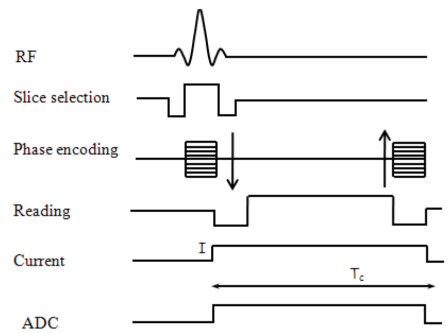

Fig. 1. Balanced-SSFP pulse sequence for MREIT

B. Simulations

The b-SSFP signal's phase variation with respect to field inhomogeneity was simulated for two uniform conductivity phantoms of TX-151 and Gelatin using the SSFP equations explained in section IIA. Their T1 and T2 values were experimentally calculated using spin echo sequence by varying TR and fixing TE and vice-versa, respectively. The experimental parameters assumed for simulations were TR of 18 ms and off-resonance phase ranging from -360^0 to +360^0. In another simulation, a phase of 50^0 and 30^0 was arbitrarily assumed to be generated in TX-151 and Gelatin due to an arbitrary current injection. The total off-resonance phase θ, is then the sum of this current injection induced phase and the field inhomogeneity phase ranging from -360^0 to +360^0. This current injection induced phase is then estimated from the b-SSFP signal phase, using the method explained in section IIA.

III. RESULTS

The experimentally calculated T1/T2 values for Gelatin and TX-151 phantoms are 500/50 ms and 800/30 ms respectively. Their conductivities were measured as 0.1 S/m and 3.1 S/m respectively. Figure 2 shows the SSFP signal phase variation against the off-resonance precession angle for the two phantoms. At the vicinity of zero off-resonance phase, the so called transition band, the phase response is almost linear. Gelatin shows higher phase sensitivity than TX-151 around the zero-phase region. Figure 3 shows the estimated current injection induced phase θ_{Bz}, for the two phantoms. It can be observed that the phase is inaccurately estimated in the regions marked by A and B. The estimated phase here was $-(360-\theta_1)$. The region A corresponds to the field inhomogeneity phase between -180^0 to (-180^0- current injection induced phase) and the regions B corresponds to the phase from (180^0-current injection induced phase) to 180^0. For other field inhomogeneity phases, the current injection induced phase was accurately estimated as θ_{Bz}.

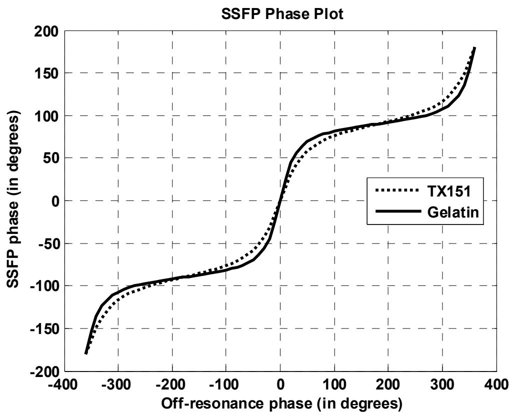

Fig. 2. Phase of balanced-SSFP signal for TX-151 and Gelatin conductivity phantoms with varying off-resonance phase angle

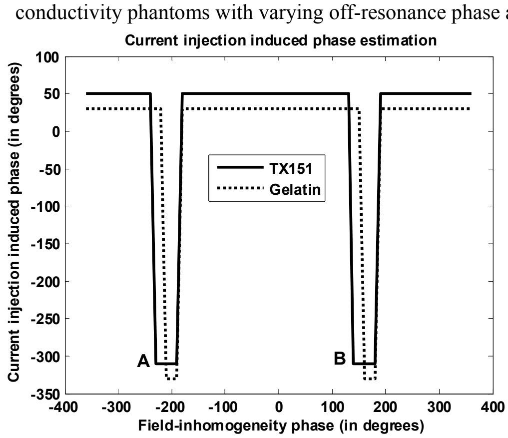

Fig. 3. Estimated current injection induced phases for TX-151 and Gelatin conductivity phantoms with varying field inhomogeneity

IV. Discussion and Conclusion

The b-SSFP pulse sequence proposed here has very high sensitivity to small deviations from resonance frequency. If MREIT experiment is performed around the resonance frequency, with negligible field inhomogeneities, it is possible to estimate the phase with high accuracy. Such phenomenon was observed for both the phantoms where the phase generated by current injection was accurately estimated. The present method explicitly assumes that T_2 value for the imaging object is already known at each pixel. Thus, its prior estimation is needed to use the method explained here. This might need another pulse sequence and hence might increase the total scan time. However, it is believed that the T_2 estimation can be done using the same b-SSFP sequence by changing experimental parameters [12]. It is possible to inject small currents of the order of 1 mA and estimate their induced phases accurately with b-SSFP sequence. This is not possible with conventional spin echo based sequence, where scan time is very large and sensitivity very small.

Future work would focus on the programming of pulse sequence in our 3T MR scanner. Our prior trials in sequence designing have suggested that the field inhomogeneity produce ringing artifacts in MR magnitude images which obscures the phase induced due to current injection. It is suggested in previous studies to use a phase cycled RF pulse for b-SSFP pulse sequence, to reduce these ringing artifacts after post processing of data [6]. We have planned to implement the b-SSFP pulse sequence based on phase cycled RF pulse and check its effectiveness in reducing the ringing artifacts. Following the pulse sequence implementation, we plan to experimentally validate the proposed b-SSFP based B_z measurement method. The b-SSFP based MREIT is expected to reduce both scan time and amplitude of injection current and human experiments would prove to be clinically feasible.

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