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Human In-vivo MR Current Density Imaging (MRCDI) Based on Optimized Multi-echo Spin Echo (MESE)

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Synopsis

MRCDI aims at imaging an externally injected current flow in the human body, and might be useful for many biomedical applications. However, the method requires very sensitive measurement of the current-induced magnetic field component $\Delta B_{z,c}$ parallel to main field. We systematically optimized MESE to determine its most efficient parameters. In one of the first human in-vivo applications of MRCDI, the optimized sequence was successfully used to image the $\Delta B_{z,c}$ distribution in the brain caused by a two-electrode montage, as confirmed by finite-element calculations of $\Delta B_{z,c}$. Further improvements will be performed to increase its robustness to field drifts.

Purpose

Imaging the current distribution injected by external electrodes in the human brain might be useful in many biomedical applications, and could, e.g. be used to reconstruct the ohmic tissue conductivities. However, in-vivo human brain MRCDI allows only for weak electrical current injection, i.e. 1-2 mA, which severely limits its sensitivity.[1] We aimed for a systematic sensitivity analysis of MESE to determine the most efficient sequence parameters for human brain imaging, and for a systematic experimental validation in phantoms. Finally, we aimed at applying the optimized sequence for in-vivo MRCDI of the human brain.

Theory and Methods

The injected current I_c creates a magnetic field $\Delta B_{z,c}$ inside brain, which is parallel to main MR field. This field causes small frequency shifts, and can be measured using MR phase images. Here, we employ MESE (Fig. 1) due to its high sensitivity, image quality, and robustness to field inhomogeneities and flow artifacts. The bipolar current is injected in synchrony with the sequence, and multiple echoes with linearly increasing current-induced phases (Fig. 2b) are acquired. $\Delta B_{z,c}$ images from each echo are calculated and optimally combined. The efficiency of MESE MRCDI η_{MESE} is given in Eq. 1,

$$\eta_{\mathbf{MESE}} = \frac{|\triangle \mathbf{B}_{\mathbf{z,c}}^{\mathbf{comb}}|}{\sqrt{\mathbf{T_{tot}}}} \sqrt{\sum_{\mathbf{n=1}}^{\mathbf{N_{echo}}} \mathbf{4}\gamma^{\mathbf{2}} \mathbf{SNR_n^2} \left[(\mathbf{T_{ES}} - \tau_\pi)\, \mathbf{n} - \mathbf{0.5}\tau_{\pi/\mathbf{2}} \right]^{\mathbf{2}}} \quad (1)$$

N_{echo}, γ, SNR_n, T_{ES}, τ_π, $\tau_{\pi/2}$, and T_{tot} are the total number of echoes, gyromagnetic ratio, signal to noise ratio of n^{th} echo, echo spacing, refocusing and excitation pulse-width, and the total measurement time, respectively. First, an experiment was performed in a saline-filled spherical phantom with relaxation parameters like brain tissue (T_1 = 1 s, T_2 = 100 ms) to determine optimized sequence parameters N_{echo} and T_{ES}. Then, an in-vivo brain experiment was conducted with a 30-years old healthy male volunteer, as approved by the local ethical board. Two MRI compatible 5x7 cm^2 rubber electrodes were placed on the head above the temporoparietal junctions of the brain. The current waveform was generated using an arbitrary waveform generator (33500B, KEYSIGHT Technologies, California, United States) and an MRI compatible transcranial current stimulator (DC-STIMULATOR PLUS, neuroConn GmbH, Germany). Two different MESE experiments with positive and negative bipolar currents were performed with field of view FOV = 256x256 mm^2, image matrix = 128x128, voxel size of 2x2x3 mm^3, N_{avg} = 1x2 (for positive and negative current injection), T_{ES} = 60 ms, dead time T_D = 1.5 s, and I_c = 1mA. In the first experiment, the number of spin echoes N_{SE} = 4 and multiple gradient echoes N_{GE} = 1 (the total number of echoes N_{echo} = $N_{SE}\times N_{GE}$; bandwidth BW = 20.2 Hz/pix) were selected. The long T_{ES} results in low-bandwidth data acquisition, causing artifacts. This can be prevented by multiple gradient echo acquisition within each T_{ES} with a very small loss of SNR. Therefore, the second experiment was repeated with N_{SE} = 4 and N_{GE} = 5 (BW = 115.2 Hz/pix). For comparison, a head model of the subject was created and the $\Delta B_{z,c}$ simulated by the finite-element method included in SIMNIBS [2] (conductivities: 0.126 S/m for white matter, 0.275 S/m for gray matter, 1.654 S/m for cerebrospinal fluid).

Results

The reconstructed $\Delta B_{z,c}$ inside phantom (Fig. 2a) is as expected for a current flow from top to bottom. The linear increase in the current-induced phase is shown for different echo spacings (Fig. 2b). The most efficient sequence parameters (Fig. 2c,d) are T_{ES} = [60-100] ms, N_{echo} = [2-4], T_D = 1.5 ms. The MR magnitude images of the human brain are shown in Fig. 3a,b, and the reconstructed $\Delta B_{z,c}$ images are depicted in Fig. 3c,d (given the occurrence of a spurious $\Delta B_{z,c}$ offset, the mean-corrected image is shown for easier interpretation). Acquiring single echo with this long T_{ES} caused image distortions, which were avoided by multiple gradient echo acquisition. The measured and simulated $\Delta B_{z,c}$ images show the same general distribution (Fig. 4), with strong field changes close to the CSF-filled and well-conducting sulci underneath the electrodes.

Discussion and Conclusion

By multi-gradient echo acquisition, the low-bandwidth artifacts are eliminated. The simulation result of the generated head model and the measurements look similar. Large magnetic field changes induced by high current density in sulcus regions are well observed near pads (Fig 3c,d). The differences between simulations and experiments may arise from rough estimation of the conductivities and anisotropy in the simulations, the spurious magnetic field induced by the current flow in the cables or electrodes, or scanner imperfections. The effect of flow, motion, and static field inhomogeneities should also be considered. Nevertheless, this study demonstrates a successful initial measurement of $\Delta B_{z,c}$ for in-vivo MRCDI.

Acknowledgements

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Figures

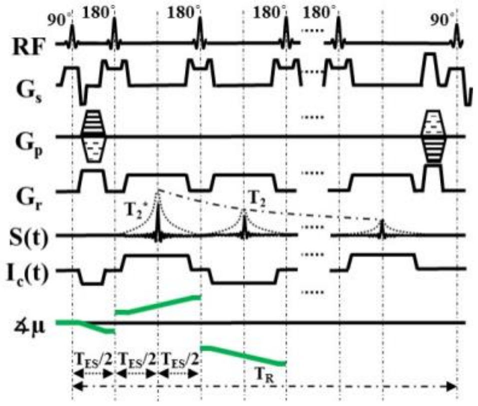

Figure 1. Diagram of the MESE pulse sequence with equal and symmetric echo spacing. The sequence is composed of a 90˚ excitation pulse preceding repetitive 180˚ refocusing pulses, so that multiple echoes are created. Crusher gradients are used to preserve only the desired echo pathways. At the end of the sequence, phase encoding rewinder and spoiler gradients are employed to eliminate the remaining transverse magnetization. The injected bipolar electrical current is synchronized with radio frequency (RF) pulses, so that the phase of the continuous complex transverse magnetization (∡μ) increases linearly over time.

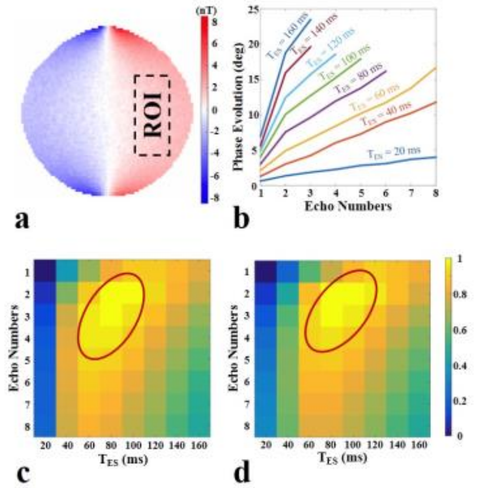

Figure 2. MESE simulation and measurements in the phantom. (a) An example of combined $\Delta B_{z,c}$ measurement for I_c = 0.5 mA. The region of interest (ROI) used to calculate the efficiency is shown by the dashed lines. (b) Measured phase evolution. (c) Simulated efficiency. (d) Measured efficiency. The results in (c-d) are normalized, and FOV = 300x300 mm^2, image matrix = 256x256, Δz = 5 mm, N_{slice} = 1, N_{avg} = 1, N_{echo} = [1–8], T_{ES} = [20–160] ms, T_D = 510 ms, T_1 = 1.1 s, T_2 = 100 ms, T_2^* = 50 ms, and I_c = 0.5 mA.

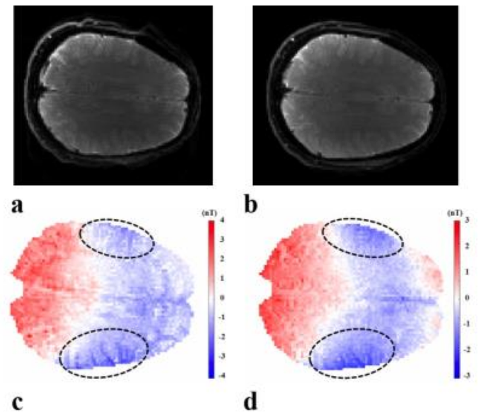

Figure 3. In-vivo MESE results in human brain. (a,b) MR magnitude images. (c,d) Reconstructed $\Delta B_{z,c}$ images. Number of spin echoes N_{SE} = 4 and number of gradient echoes N_{GE} = 1 are selected in (a,c), and N_{SE} = 4 and number of gradient echoes N_{GE} = 5 are selected in (b,d). Other parameters are field of view FOV = 256x256 mm^2, image matrix = 128x128, voxel size of 2x2x3mm^3, N_{avg} = 1x2 (for positive and negative current injection), T_{ES} = 60 ms, dead time T_D = 1.5 s, and I_c = 1mA.