A simple method to image time-varying magnetic fields inside the body using MRI.

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Target Audience: Researchers and clinicians interested in dosimetry techniques for RF ablation, transcranial stimulation, or any other application requiring oscillating magnetic fields inside the body.

Introduction: The aim of the present work is to present proof of concept of a simple method to image *time-varying* magnetic fields induced in the body by an electromagnetic probe (therapeutic or otherwise) that is driven by an arbitrary known waveform. Static magnetic field imaging has been demonstrated by observing phase shifts in the signal resulting from changes in echo time (*1-4*), but mapping oscillating fields remains limited to the resonant frequency of the scanner (*5*) (*6*). The approach presented in this work considers this magnetic field as a separable function in time and space (i.e., $\vec{B}(\vec{r},t) = \vec{B}(\vec{r})\vec{B}(t)$). The method allows mapping the *spatial* distribution of all three Cartesian components of a magnetic vector field that varies with an arbitrary known function, provided that its frequency content is significantly below the resonant frequency of the system.

Methods: All images were collected on a 7 Tesla scanner (7/310, Agilent Technologies, Walnut Creek, CA) using a birdcage RF coil for excitation and acquisition. The first experiment was to verify our ability to image the magnetic field projections over a range of oscillating magnetic field intensities and frequencies. An angled figure-eight shape coil (loop diameter = 3cm, angle ~ 130 degrees) was constructed from magnet wire in order to mimic a TMS coil, and placed on a spherical phantom (diameter = 5cm) filled with water. The phantom-coil assembly was placed inside the scanner, oriented so that the bulk of the magnetic field produced by the coil was aligned with the main magnetic field of the scanner. The coil was driven by an arbitrary function generator (HP 33120A, Palo Alto, CA). Ten slices were collected along the axial plane (or XY-plane in the scanner's frame of reference) using a standard gradient echo sequence (TR = 116 ms, TE = 10 ms, flip angle = 20 deg, FOV = 8 cm, slice thickness = 2 mm, matrix size = 128 x 128, RF and gradient spoiling were used to remove spurious magnetization). A sine wave consisting of 3.25 cycles was transmitted through the probe immediately after the RF excitation pulse, but before the spatial encoding gradients of the pulse sequence. The phase gain ($\Delta\phi$) induced in the image at location **r** by *this specific waveform* can be shown to be given by $\Delta\phi = 2\pi\gamma B_z(\vec{r})\frac{1}{\omega}$. Where $B_z(r)$ is the z-component of the magnetic field at position **r**. The peak-to-peak voltage and frequency of the waveform were varied in order to verify the predicted phase gains (0, 1, 5, 10 and 14 V; and 0.5, 1, 5, 10, 50, 100 kHz). The images were reconstructed and the phase gain relative to a reference image (without the pulse) was calculated and analyzed as a function of the input voltage and frequency.

A second experiment was carried out to test our ability to reconstruct the three Cartesian components of the oscillating magnetic vector field. A small phantom was constructed by filling a small bottle with gelatin and introducing a thin hollow tube diagonally inside the gelatin, in order to give the phantom an asymmetric structure. A figure-eight probe (6 turns on each loop, 2 cm diameter loops) was attached to the phantom and the whole assembly was placed inside the scanner. The object was imaged at seven different orientations using a gradient echo sequence (TR = 60 ms, TE = 4 ms, flip= 30 deg, FOV = 6.4 cm, sl thick= 1 mm, matrix size = 64 x 64 x 50, RF and gradient spoiling). The phantom/probe assembly was imaged at each position twice: with and without current running through the probe immediately after the excitation RF pulse. This pulse lasted 1.5 ms and consisted of three cycles (i.e., frequency = 2 KHz) of a DC-shifted cosine wave oscillating between 0 and 1 Volts. The images were subsequently reconstructed and the phase difference between the complex images with and without the magnetic pulse was computed at each position. The projection of the magnetic field onto the Z-axis (B_z) was computed from these phase differences and from the integral of the applied magnetic pulse. The 3D vector field was reconstructed from it projections by the method previously published in (*4*).

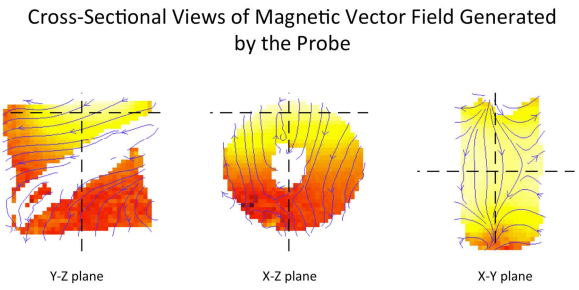

Figure 1: log-scale map of the spatial distribution of the magnetic field. Streamline arrows indicate field direction.

Results: The first experiment confirmed the linear dependence of the phase accrued due to the probe's magnetic field on the frequency and amplitude of the field. There was a strong linear relationship between increasing phase gain and the frequency (r> 0.96) and peak-to-peak voltage of the pulses (r> 0.82). The second experiment yielded all three Cartesian components of the magnetic vector field. Figure 1 displays three intersecting orthogonal views of that magnetic vector field's magnitude on a log scale to capture the full dynamic range. Streamlines are overlaid on top of the magnitude images to indicate the direction of the magnetic field that resulted from the projection process (Note that the empty space in the middle is due to the partially filled hollow tube).

Discussion: We have presented a simple method to map time-varying magnetic fields inside tissue using MRI. This method is quite versatile in that it allows for imaging the spatial distribution of a broad spectrum of frequencies and only requires that the device in question be MR compatible and can be synchronized with the MR scanner, which can be usually be achieved via TTL lines. The main constraints of the method are that (1) it requires knowledge of the temporal characteristics of the magnetic field under scrutiny, and tight control of its timing; (2) it can only be used for low field amplitudes (in the order of a few Gauss) so that the phase gain can be kept within 0 and 2PI □□□□□□□□, which means that in most cases we must image a scaled down version of the desired field; (3) the frequency content of the waveform must be below the Larmor frequency of the tissue's spins.

Conclusion: The proposed technique yields an accurate spatial map of the magnetic vector field in the target region, given its time course.

References: (1) Chen et al:Magn Reson Med 41, 1206 (1999). (2) Bohning et al.: Neuroreport 8, 2535 (1997). (3)Jezzard et al: Magnetic Resonance in Medicine 34, 65 (1995). (4) Hernandez-Garcia, et al: Neuroimage 36, 1171 (2007). (5) Sacolick, F. et al: Magnetic Resonance in Medicine 63, 1315 (2010). (6) Halpern-Manners, NW. et al: PNAS 107, 8519 (2010).