Oscillating Magnetic Field Mapping using MRI

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INTRODUCTION: We present a new method to estimate oscillating magnetic fields in the Kilohertz (KHz) frequency range using magnetic resonance imaging. Our target applications are to map the effects of "metamaterial" lenses in transcranial magnetic stimulation (TMS). The recently developed "metamaterials" can be used to focus electromagnetic fields onto regions that are smaller than the field's wavelength, previously considered impossible [2]. Oscillating magnetic fields have also been proven useful for targeted drug delivery application [3]. Externally focused magnetic fields of known intensity could possibly improve targeting capabilities of such procedures. Our strategy consists of constructing a forward model for the MRI signal produced by a known object in the presence of an oscillating magnetic field of known temporal characteristics but unknown spatial distribution. We use iterative techniques to estimate the distribution of the field by comparing the measured signal to the signal generated by a forward model. Here, we develop the theoretical framework and illustrate our method by numerical simulations and experimental data.

METHODS: *Simulations*: We simulated a MR image of the Shepp-Logan phantom (32 x 32, FOV = 60 x 60 mm) in the presence of a sinusoidal magnetic field generated by a circular coil (diameter of 4cm, with a current of 5 mA flowing through it) positioned parallel to the slice of interest. For simplicity, the field was simulated using the Biot-Savart equation assuming no additional fields created by charge build up. The field was synchronized with the data acquisition phase of the pulse sequence. Thus, the MR signal *s (t)* is given by

$$s'(t) = \int_{r\in volume} \rho(r)\, e^{-i\,k(t)r}\, e^{-i\frac{\gamma}{2\pi}\int_0^t B(r)\sin(\omega\tau)d\tau}\, dr$$

where ω is the frequency of sinusoidal current, *B(r)* is the magnetic flux density at location r, and ρ*(r)* is the spin density at that location**.** Gaussian noise was added to the modeled signal so that the SNR of the reconstructed images varied from 26 to 32 dB. In order to solve for the magnetic field, the following cost function was minimized using a non-linear least squares algorithm.

$$\hat{B}(r) = \min_{B(r)} \|s'(t) - \hat{s}(t)\| + \beta R(r)$$

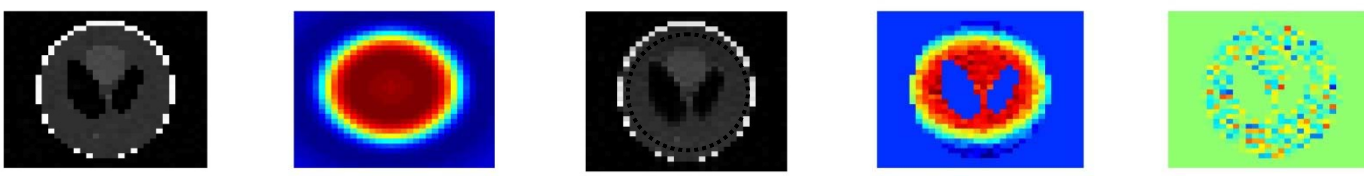
where *R(r)* is a spatial roughness penalty, as in [4]. This penalty consisted of the first order derivative of the map and was weighted by a factor β. The effect of this regularization term on the residual error was investigated by varying β from 0 to 0.1 at a SNR of 20dB

Experimental Data: All images were acquired on a Varian (Palo Alto, CA) 7T scanner. A figure-eight (Diameter=3 cm, 15 turns) probe was placed adjacent to a spherical phantom (diameter = 5 cm) as indicated in the figure. The assembly was placed in the scanner such that the axis of the coil was approximately parallel to the B_0 magnetic field of the scanner. Two images were acquired using a gradient-echo sequence (TR: 350 ms, TE: 0.64 ms, flip angle = 40 degs, BW = 50 KHz, Matrix Size = 32 x 32, Slice Thickness = 2 mm, FOV = 60 x 60 mm). We acquired images of the phantom in axial, coronal and sagittal planes both with and without the magnetic field of interest being applied. A current of 0.5 mA at 1 KHz was circulated through the probe during the collection of one image. No current was present for the second image acquisition. TTL triggers from the MR master console were used to synchronize the applied field with the read encode gradient.

RESULTS: *Simulations*: Figure **1a** and **b** show the simulated true object (SNR = 30 dB) and magnetic field map Figure **1c** shows the simulated image in the presence of the oscillating field (dashed lines indicate location of probe above the object). Figure **1d** shows the estimated magnetic field over the object. Figure **1e** shows the difference between true and estimated magnetic field values over the object (RMSE = 0.044 gauss). True and estimated field maps are displayed on a scale of [-0.2 to 1.6] gauss and the error map is displayed on a scale of [-0.3 to 0.3]. Figure 2 indicates the sensitivity of the method to the SNR of the system.

Fig.1

a b c d e



RMSE = 0.044 gauss

Figure **2a** shows plot of the RMSE v/s the SNR of the image. Trend shows that the RMSE decreases as the SNR of the image increases. The field map was estimated at $\beta = 2^{-4}$ after 75 iterations. Figure **2b** shows a plot of the RMSE v/s β. At SNR of 20dB, β = 0.05 was optimal after 20 iterations. We observed that the optimal value of β varies for images with different SNR. However, if the number of iterations were large enough (>50) regularization made little difference to the RMSE.

Experimental: Figure **3** shows coronal, sagittal and axial planes of the magnetic field estimated over the phantom from the experimental data respectively. The position of the coil relative to each image is indicated by the black ovals. Also the relative location between the three sections is illustrated on the coronal section by the black boxes. The distribution of these magnetic fields is consistent with the expected fields generated by two wire loops.

Figure 2a

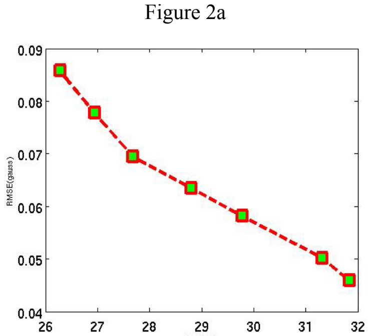


Figure 2b

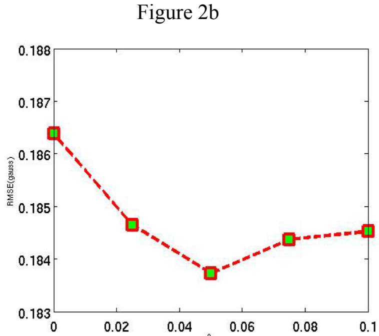
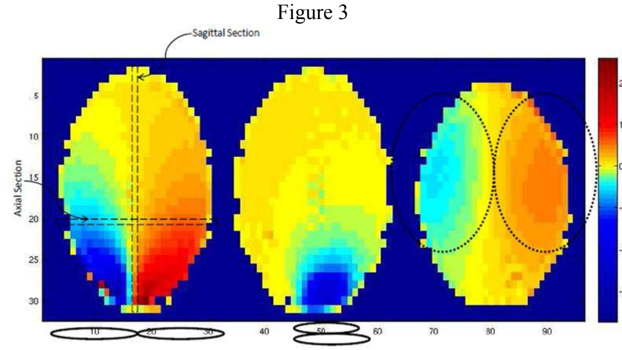


Figure 3



Discussions and Conclusion: We have demonstrated a method to estimate an oscillating magnetic field map in the KHz range in simulated and experimental data. The model for estimation assumes that static susceptibility artifacts due to the presence of the coil are minimized by fast imaging techniques and short echo times. It is also important to note that the field estimates we obtain are for the z-component of the magnetic field. We expect that the other components can be estimated using the procedure discussed in [1]. Experimentally, the amplitude of the field must be chosen such that the effect is large enough to be observable but not large enough to cause the MR signal to dephase completely. Moreover, this method can only estimate the field map over a region where there is some sample present.

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References: [1] Hernandez-Garcia, NeuroImage 36 (2007) 1171–1178. [2] Pendry. Physical Review Letters. 0031-9007/00/85(18)/3966(4), 2000. [3] Edelman 1985, [4] Funai IEEE Transactions on Medical Imaging, VOL. 27, NO. 10, (2008).