

Is 4.0 T MRI advantageous for current source localization of weak magnetic fields?

K.D. Brewer^{a,b}, B.L. Osberg^{a,b}, J.A. Rioux^a, R.C.N. D'Arcy^{a,c,d},
G. Stroink^b, S.D. Beyea^{a,b,d,*}

^a *Institute for Biodiagnostics (Atlantic), National Research Council of Canada, Halifax, NS, Canada*

^b *Department of Physics, Dalhousie University, Halifax, NS, Canada*

^c *Department of Psychology, Dalhousie University, Halifax, NS, Canada*

^d *Department of Radiology, Dalhousie University, Halifax, NS, Canada*

Abstract. Recently, several methods have been proposed for measuring the magnetic fields produced by weak electric currents using magnetic resonance imaging (MRI), including a theoretically described technique for source localization of current dipoles using signal phase. We have investigated whether these results can be replicated experimentally using 4.0 Tesla (T) MRI with a dipole phantom. While the use of high magnetic fields is advantageous in conventional functional MRI, it is unclear, *a priori*, whether MR current source localization also benefits. This study presents the technical challenges in MRI current source imaging and quantifies sensitivity using a phantom containing a current dipole. Crown Copyright © 2007 Published by Elsevier B.V. All rights reserved.

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1. Introduction

Functional magnetic resonance imaging (fMRI) is a common technique for localizing brain activation. FMRI does not directly quantify neural activity, but instead measures the haemodynamic response that occurs as a result of this activity, a

* Corresponding author. Neuroimaging Research Lab, Suite 3900, QEII Health Sciences Centre, 1796 Summer St. Halifax, NS, Canada B3H 3A7. Tel.: +1 902 473 1850; fax: +1 902 473 1851.

E-mail address: steven.beyea@nrc-cnrc.gc.ca (S.D. Beyea).

property referred to as the Blood Oxygen Level Dependent (BOLD) effect [1]. Conversely, electroencephalography (EEG) and magnetoencephalography (MEG) directly monitor the electrical activity of neurons [2]. In order to fully characterize brain function, methods that better link MRI to EEG/MEG are needed.

A possible solution is to use signal changes in MRI to measure electric or magnetic fields arising from neural activity [3,4]. MRI current source localization could complement EEG/MEG spatial localization by providing valuable information that can be used to constrain inverse solutions. Typically, the localization of EEG/MEG sources requires MRI data for anatomical information. Localization of current sources using MRI would provide both functional and anatomical information in a single session.

Recently, a number of MRI methods have been proposed for directly imaging the weak magnetic fields produced by neural activity [3,4]. These methods have been validated in phantom models; however, their reported detection in human functional imaging is controversial because the magnetic fields created by neurons are extremely weak (and therefore difficult to measure) compared to those used in MRI. Furthermore, these studies have been performed at “clinical” magnetic field strengths (i.e. ≤ 3 Tesla (T)). It is well established that conventional BOLD fMRI exhibits increased sensitivity at high magnetic fields [1]. We will therefore address whether MRI of weak magnetic fields also shows evidence of increased sensitivity at a magnetic field strength of 4.0 T.

2. Experimental design

2.1. Phantom design

All experiments were carried out on a phantom containing a 1-cm copper wire twisted pair current dipole (Fig. 1A), mounted to a solid non-conducting rod made from Ultem™ polymer (GE Plastics Inc.). Ultem™ has the same magnetic susceptibility as water, which reduces image artifact. The dipole was placed in a 184-mm-diameter flask containing a 0.5% NaCl solution and a 1.3% agar gelling agent (BDH Laboratory Chemicals Div.) to reduce the flow due to hydrodynamic propulsion. Current was supplied using a function generator (Wavetek, Model 190). A magnetic fluxgate sensor (Automation Industries Inc) was used to measure the radial component of the magnetic fields near the outer surface of

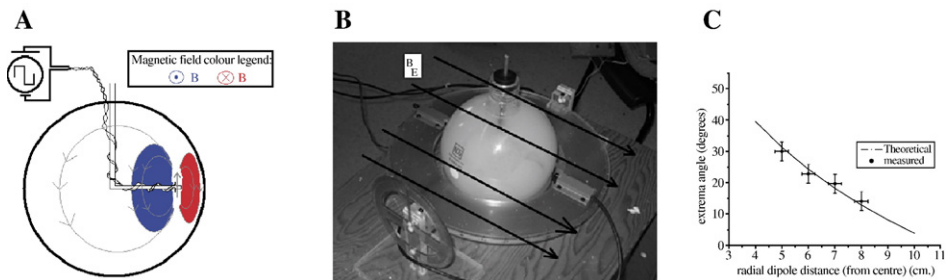


Fig. 1. (A) Schematic of the dipole phantom. (B) Flask containing the dipole, placed in a gantry system containing the magnetic flux gate apparatus. (C) Plot of extrema angle versus dipole radial distance from the center of the flask, as measured using magnetic fluxgate measurements 2 cm from the surface of the phantom.

the flask due to the current dipole (Fig. 1B and C). This verified that the dipole phantom produced the characteristic field patterns of a dipole.

2.2. Magnetic resonance imaging experiments

The basis for the detection method [3,4], is that the MRI signal phase depends upon the signal evolution time (TE) and the total local magnetic field; as described by Eq. (1)

$$\theta = \gamma \times (B_0 + \Delta B) \times TE, \quad (1)$$

where γ is the nuclear gyromagnetic ratio, B_0 is the main magnetic field, and ΔB is the component of the magnetic field created by the dipole that is parallel to the main field.

As shown in Eq. (1), the phase sensitivity (and therefore sensitivity to the dipole) will increase linearly with TE, at the expense of increased image artifact and decreased signal-to-noise ratio (SNR). Initial validation experiments showed that, in our dipole phantom model, the optimal TE (based upon artifact level, etc.) was 15–20 ms.

Imaging data sets were collected at varying current amplitudes, corresponding to differing dipole strengths. Currents through the current dipole of ± 2 mA, $67 \mu\text{A}$, $30 \mu\text{A}$, $18 \mu\text{A}$, $5 \mu\text{A}$, $2 \mu\text{A}$ and $1 \mu\text{A}$ were used. The dipole waveform was a 0.179 Hz sinusoidal square wave, with the MRI triggered to scan every time the current changed polarity. MRI data was collected using a conventional 2-shot EPI sequence with TR per volume = 2800 ms, TE = 17 ms (or 20 ms for currents $\leq 5 \mu\text{A}$), $8/2$ mm thick slice/gap, 240 mm field-of-view (64×64 matrix), and 35° flip angle. Images were collected for a total of 200–800 repetitions (800 was used for the lowest currents).

The MRI data was analyzed using IDL 6.1 (Research Systems Inc.). Phase maps were generated for each image slice, with the phase calculated as the arctangent of the complex MRI data. Phase maps underwent phase unwrapping and pair-wise subtraction. “Activation” maps were then generated by performing an analysis which looked for clusters of pixels whose phase changes correlated with the time course of the current dipole amplitude. MRI measurements were acquired using a 4.0 T Varian/Oxford whole-body scanner, and a TEM volume RF coil (Bioengineering Inc.).

3. Results and discussion

3.1. Sensitivity limits using 4.0 T MRI

For a typical physiological current dipole of approximately 10 nA m [2] at a detection distance of one imaging pixel from the dipole ($\approx 4 \text{ mm}$), the magnetic field is roughly 10^{-8} to 10^{-9} T. Detection of an approximately 10^{-7} T magnetic field perturbation was possible in a single-trial experiment (one “positive/negative” set).

Improvements in sensitivity are obtained using correlation analysis of the repeated on/off conditions. Correlation maps at several current levels are shown in Fig. 2. Our detection threshold appeared to be $\pm 2 \mu\text{A}$ (800 polarity reversals), which corresponds to a dipole fluctuation of approximately $\pm 20 \text{ nA m}$. This limit could be lowered using filtering, as the

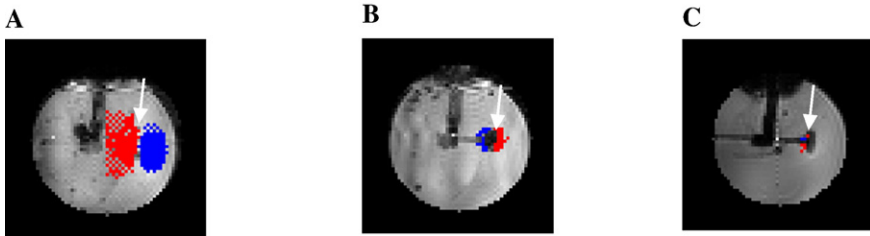


Fig. 2. Correlation plots (over-laid on a raw MRI image of the phantom), showing detection of a current dipole created at a current level of (A) ± 2 mA, (B) ± 67 μ A and (C) ± 2 μ A. The side showing positive or negative correlation varies based upon the initial direction of the electric current. The white arrow denotes the location of the dipole. The 4-T magnetic field is orthogonal to the plane of the image.

function generator (located outside the magnet room) transmitted electronic noise into the RF coil via the conducting wire. Further experiments on this dipole phantom are therefore necessary to determine the ultimate detection limit at 4.0 T.

3.2. Sensitivity of 1.5 T versus 4.0 T MRI

It is well established that image artifact levels, for equivalent TE values, are decreased when using lower field (e.g. 1.5 T) MRI. Typically, for 1.5 T versus 4.0 T, it would be possible to double the TE during an acquisition while maintaining an equivalent level of image artifact. In theory, therefore, one should be able to double the phase sensitivity of such measurements by using lower fields. However, the SNR scales with magnetic field strength as $B^{3/2}$ [5]. It is therefore not immediately apparent which of these factors will lead to the best phase contrast-to-noise ratio (CNR).

In a representative 4.0 T data set, we found a maximal phase offset of 0.27 rad with a SNR of 9.1. By calculating the necessary signal change for that phase offset, and propagating the error, we found a phase CNR of 1.82.

Assuming that the RF coil sensitivity and all sources of noise remain the same at the lower field, we then calculated the phase CNR that would have been obtained in that image at 1.5 T when using twice the TE utilized in the 4.0 T experiment. While the absolute phase contrast would increase to 0.54 rad, the SNR would decrease to 4.22, leading to a phase CNR of 1.16. It would therefore appear that the use of 4.0 T MRI is in fact advantageous for mapping of weak dipole fields using phase contrast imaging, even when using half the TE of experiments performed at 1.5 T.

4. Conclusions

The use of 4.0 T MRI technique permits the detection of weak current dipoles, with a detection limit in the present experiments of ± 20 nA m. Smaller fields may be detectable through further improvements in SNR and artifact level.

While the use of high field MRI limits data acquisition using long detection times, because of increased image artifact/distortion, initial indications are that the decrease in

potential phase contrast is more than offset by the increase in SNR, leading to an improvement in phase CNR. We therefore conclude that further experiments in dipole phantoms or humans will benefit from the use of high field MRI.

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