Vector magnetic field mapping of a Transcranial Magnetic Stimulation coil using Magnetic Resonance Imaging: *in vitro* and *in vivo* experiments

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Abstract— **Non-invasive functional brain techniques are becoming increasingly important to foster important advances in neurosciences. Transcranial magnetic stimulation (TMS) is one possibility whose applications have been spreading to several areas of neuroscience. By producing high intensity magnetic flux changes TMS is able to depolarize neurons thus eliciting brain or nerve responses. However, the ability to focus the magnetic pulse to small areas is limited and the stimulus is usually restricted to cortical areas. To further improve TMS performance, new coil design will be necessary as well as methods to quickly map the 3D produced magnetic field vector. Magnetic resonance imaging (MRI) was used in this work to map the magnetic field produced by a TMS butterfly coil. Measurements were performed *in vitro* and *in vivo*. The reliability of the results suggests the use of MRI to produce magnetic field mapping, in conjunction with other techniques, to develop TMS coils for different purposes.**

Keywords— **TMS, MRI, Vector Field, 3D Mapping.**

I. INTRODUCTION

Transcranial magnetic stimulation [1] is a technique that uses short (approximately 300μs duration) and high intensity (approximately 2T) magnetic pulses to depolarize regions of the brain cortex or nerves. Such stimulus produces a behavioral response which is dependent on the site of neuronal population being evoked. For instance, if applied to the motor cortex, an involuntary contraction of certain group muscles will be evoked, which may be observed by the measurement of the Motor Evoked Potential – MEP.

For that purpose, different types of stimulation coils are used. The most common ones are the circular coil, usually with a small diameter for children and large diameters for adults, and the butterfly configuration, which is composed of two flat coils carrying the same current, with parallel winding, but oriented in opposite-direction. Figure 1 shows an X-Ray image of a butterfly coil.

The usual way to map magnetic fields produced by these coils, is to set a magnetic sensor in every point to allow the construction of the field space distribution. This procedure is time consuming and if not automatic, very tedious. On the other hand, MRI phase images can be an attractive alternative.

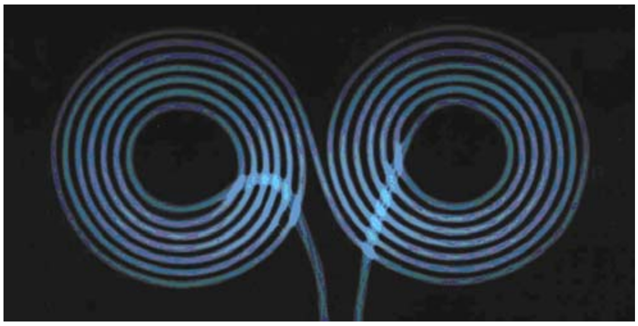

Fig. 1 X-ray image of a butterfly coil showing the orientation of the windings made of thick copper wire. The scale is given in millimeters.

Several methods have been proposed for magnetic field mapping using Magnetic Resonance Imaging (MRI) [2-4]. In gradient recalled echo (GRE) experiments, the spins are only partially refocused at time to echo (TE) by the reversed readout gradient. The spin phase in each position ($\vec{r}$) is linear to the axial component of the magnetic field inhomogeneity (ΔB_z). The phase accumulated (ϕ) at TE follows the expression:

$$\phi(\vec{r}) = \gamma \cdot \Delta B_z(\vec{r}) \cdot TE \ ,$$

where γ is the proton-gyromagnetic ratio.

In order to map the magnetic field produced by coils, Tomasi and Panepucci [2] obtained two GRE images without/with current in the coils using the same TE. The accumulated phase in each case is:

$$\phi_{without}(\vec{r}) = \gamma \cdot \left[\Delta B_z(\vec{r})\right] \cdot TE$$

$$\phi_{with}(\vec{r}) = \gamma \cdot \left[\Delta B_z(\vec{r}) + Bcoil_z(\vec{r})\right] \cdot TE$$

From the difference between both phase images the axial magnetic field distribution generated by the coil ($Bcoil_z$) can be obtained. Nevertheless the discrete and wrapped nature of phase images complicates the reconstruction of the magnetic map. The phases have been wrapped into the range [-π, π] and showed discontinuities over each transition. To find the correct field map, it is necessary to apply an algorithm known as “unwrapping”, which eliminates these discontinuities [6,7].

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In the MRI method, only the component parallel to the static magnetic field (B_0) of the scanner will be measured. This drawback can be solved by rotating the phantom and the coil related to B_0, and these set of images allow the reconstruction of a 3D vector magnetic field. In this work we obtained the magnetic field vector map generated by a TMS coil using in vitro and in vivo MRI experiments.

II. Materials and methods

For *in vitro* experiments two phantoms were prepared containing a solution that simulates the magnetic resonance conditions (relaxation times and conductance) of an average human tissue. The solution was composed by: 3.6g of NaCl and 1.9555g of $CuSO_4 \cdot 5H_2O$ per liter of H_2O [7]. One phantom was a 13 liter tank (13x25x40 cm^3) and the other one a real size plastic head.

All the images were acquired at 3 T (Philips Achieva, The Netherlands) MRI scanner using Gradient Recalled Echo sequences and radiofrequency transmit/receive whole body coil suitable for the experiments. The time to echo (TE=4.6 ms) was chosen for high signal-to-noise ratio, eddy currents artifacts reduction and enough magnetic field sensitivity at the phase image. The imaging parameters included: Repetition time 360 ms, echo time 4.6 ms, Flip angle 45°, FOV 25 x 25 cm, Acquisition matrix 512 x 512, Slice thickness of 2 mm and 1 mm of gap. All images covered a region of 8 cm beyond the surface of the TMS coil.

A butterfly coil from a TMS device (NeuroSoft, Russia), model Neuro-MS was used. A circuit composed by a resistor (10Ω and 50W) connected in series to the butterfly coil and a 12V lead-acid automotive battery was used, the resistance of the TMS coil is 0.3Ω, and a 1.12A current passed trough the circuit. In order to know whether the MRI method would be sensitive enough, a fluxgate magnetometer was used to obtain the field value around the coil, and it was on the order of 20 μT, enough to be measured by the method described.

It was impossible to use the TMS current source to directly feed the coil, even though it produces only high intensity pulses, around 1.5T, but with short duration, around 300μs long, and it is necessary a greater pulse duration so that all the slices can be obtained in proper conditions.

Images were obtained with the coil turned on and off, in 3 orthogonal directions using both phantoms. Thus, given a specific pixel of a slice positioned in a determined distance from the surface of the coil, the information of this point represents one of the 3 components of the magnetic field vector in that position. In order to obtain the other 2 components, it is necessary to find the information of the same point in the others 2 slices. A volumetric co-registration process was conducted using the magnitude images and three markers, which consisted of vitamin E capsules fixed on the surface of the butterfly coil and on the head phantom.

All phase images were unwrapped using a function implemented in Matlab. After the phase difference, the vector map was created for every voxel of the volume.

In the last experiment, *in vivo* images of a volunteer (male, 20 years old) were acquired using the same mentioned parameters. A handmade TMS coil, similar to the one used in the previous experiment, was rigidly positioned in the top of the head (vertex) and only the longitudinal component of TMS coil magnetic field was determined. Informed written consent was obtained before the examination.

III. Results and analysis

To compare the three magnetic field component, we first performed the co-registration that consists of rotation in the six degrees of freedom, referenced by the three landmarks, and segmentation such that the volume matrices have the same dimension.

Figure 2 shows the MRIs associated to the measurements in the three directions and the respective field map, after the use of the unwrapping algorithm.

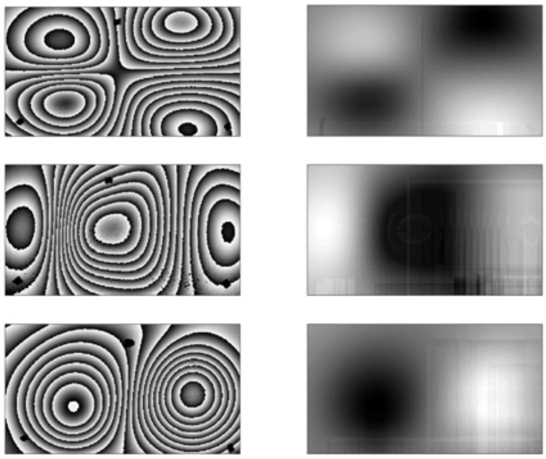

Fig. 2 Difference phase contrast images due to the magnetic field produced by a butterfly TMS coil (left), and its associated field maps (right). From top to bottom the mapping in the x, y and z axes of the TMS coil. The scale is given in millimeters.

We also mapped the field inside a head phantom and as this phantom is filled with the same solution as the tank, the images obtained were very similar (Fig. 3).

Finally, we acquired an image of a subject with the TMS coil positioned at the top of his head. A dipolar magnetic pattern can be seen, although in the human head there is less variation than in the phantom. *In vivo* measurements do not allow the measurement in the three axes, thus only one direction was possible. However this information is also relevant, because of deep stimulation, the z contribution is more intense than the other components and it is the dominant in the stimulation process.

Figure 3 shows the phase images of the head phantom and the brain of a subject, both submitted to the magnetic field generated by a butterfly coil.

Fig. 3. Phase images of head phantom filled with $CuSO_4$ solution (upper) and subject brain (bottom) submitted to a magnetic field produced by a butterfly coil positioned at the vertex (VC) position. The scale is given in millimeters.

Using the previous information a magnetic vector map was constructed, with the phantom images and overlapped onto the subject's image (Fig. 4). Interestingly, this map allows an estimation of which areas of the brain are being stimulated by the vector magnetic field. In other words, it is possible to know not only the intensity, but also the orientation of the field. This second piece of information can be as relevant as the first, since the stimulation will be strongly dependent on the ability of the magnetic field to couple with the brain fibers and neurons.

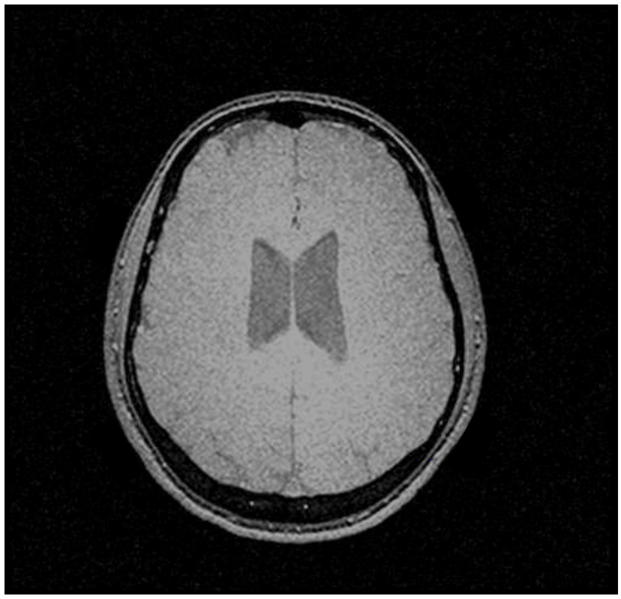

Fig. 4 Magnetic field vector map overlap in the subject head MRI imaging. The butterfly coil is positioned at the vertex position. The image quality is influenced by the RF coil used. The scale is given in millimeters.

IV. CONCLUSIONS

The vector mapping of the magnetic field produced by a butterfly coil was performed using phase contrast MRI, conducted in phantoms and in a volunteer. This information could increase the precision of the application TMS. The magnetic field mapping by phase contrast MRI demonstrated that it is a good alternative to measure complex fields, and also to help in the development of more sophisticated coil arrangement aiming the stimulation of deeper structures of the brain.

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