Clinical Neurophysiology 114 (2003) 2204–2209
www.elsevier.com/locate/clinph

Modeling the effects of electrical conductivity of the head on the induced electric field in the brain during magnetic stimulation

Kent Davey[a,*], Charles M. Epstein[b], Mark S. George[c], Daryl E. Bohning[d]

[a]University of Texas, Pickle Research Center, Center for Electromechanics, J. J. Pickle Research Campus, 10100 Burnet Road, EME 133, Austin, TX 78758, USA
[b]Department of Neurology, Emory University, School of Medicine, Atlanta, GA, USA
[c]Department of Psychiatry, Medical School of South Carolina, 67 President Street, Box 25086, Institute of Psychiatry, Room 502, North Charleston, SC 29425, USA
[d]Department of Radiology, Medical School of South Carolina, 67 President Street, Box 25086, Institute of Psychiatry, Room 502, North Charleston, SC 29425, USA

Accepted 18 June 2003

Abstract

Objective: The objective of this document is to quantify the effect of changing conductivity within the brain in transcranial magnetic stimulation.

Methods: Extreme examples of white and grey matter distributions as well as cerebral spinal fluid are analyzed with numerical boundary element methods to show that the induced E fields for these various distributions vary little from the homogeneous case.

Results: Models representative of the brain that demarcate regions of white matter and grey matter add an unnecessary level of complexity to the design and analysis of magnetic stimulators. The induced E field varies little between a precise model with exact placement of white and grey matter from that of its homogeneous counterpart. The E field will increase in white matter, and decrease in grey, but the variation is small. The contour integral of the E field around a closed path is dictated by the flux change through that contour.

Discussion: The maximum value of the variation of the electric field between a fully homogeneous medium, and one filled with different conductivity media is 1/2 the conductivity ratio of the media involved. Neuronal stimulation is more likely at the interface between dissimilar mediums, the greatest being between white matter and cerebral spinal fluid. The interface location where no normal electric field exists will witness a localized electric field 51% greater than the homogeneous E field on the white matter side of that interface. White–grey matter interfaces will have a maximum localized increase in the E field 22.9% greater than the homogeneous case.

Conclusions: Variations in neural intracellular potential during a magnetic stimulation pulse will be small among patients. The most efficient modeling will follow by assuming the medium homogeneous, and noting that perturbations from this result will exist.

1. Background

Magnetic stimulation of the human brain is now common practice both as a research and a diagnostic tool, and is being increasingly employed in the treatment of depression (Epstein and Davey, 2002). In TMS, a time varying magnetic field held near the scalp induces electrical currents in brain. An important question for researchers in this arena is determining exactly where in the brain TMS induces electrical activity, and whether this shifts as a function of differences in conductivity and organization of grey matter, white matter and CSF (Bohning et al., 1997, 2001; Analylist et al., submitted for publication; Wassermann et al., 1996). A number of effective homogeneous models of the TMS magnetic field have been proposed (Roth and Basser, 1990; Ueno et al., 1988, 1990). Liu and Ueno (1998) proposed that when current flow from a lower conductivity region to a higher one, the interface acts as a virtual cathode. An analogy is then drawn to infer the similarities between conventional electric stimulation and magnetic.

This manuscript would support that inference and underscore the fact that at the point that positive ions are driven into a nerve cell, its intracellular potential will rise, and if the rise is sufficient, an action potential results. The electric field is larger on the lower conductivity region.

* Corresponding author. Tel.: +1-512-232-1603; fax: +1-512-471-0781.
E-mail address: k.davey@mail.utexas.edu (K. Davey).

1388-2457/$30.00 © 2003 Published by Elsevier Ireland Ltd. on behalf of International Federation of Clinical Neurophysiology.
doi:10.1016/S1388-2457(03)00240-2

The nerve cell cannot distinguish whether the rise occurred because of a rapidly changing magnetic field or an imposed electric field. The inner skull boundary condition insures that the normal component of current density is essentially zero on that boundary.

2. Methods

Several models are tested to determine how the inclusion of volume conductivity differences affects the induced electric field. The analyses are performed using numerical boundary element methods (Zheng, 1997). In this technique equivalent surface sources are sought to satisfy Ampere's and Faraday's law in the medium and the boundary conditions on the surface. Both surface current and surface charge are required on every interface. The induced E field is proportional to frequency; the frequency is held fixed at 5740 Hz in all calculations as are the amp-turns. The brain is represented as a 7.5 cm radius sphere. Due to symmetry only 1/4 of both the C core stimulator and the target is modeled.

After each computation, the induced E field was determined throughout the brain twice, the second time with a homogeneous conductivity throughout. The numerical solver has the ability to compute and display the E field at a number of preset densities. Although the density was held fixed throughout this study, the number of comparison points changes slightly due to the program's need to vary the location of the computed E field as discrete region volumes change.

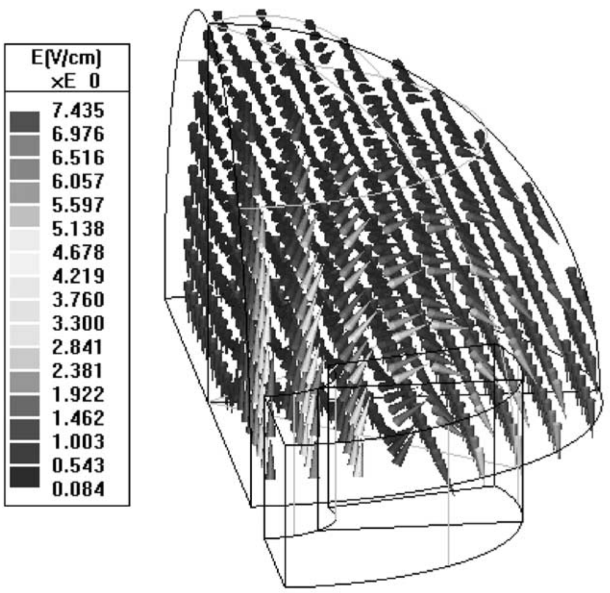

Fig. 1. Homogeneous model of the brain with conductivity 0.75 S/m throughout.

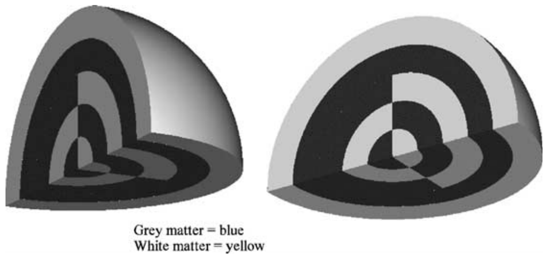

Fig. 2. Concentric sphere distribution of grey and white matter.

3. Results

3.1. Homogeneous model

Consider first the homogeneous model shown in Fig. 1. The brain is depicted as a homogeneous sphere with radius 7.5 cm. The outer surface corresponds to the skull. 1/4 of the problem is worked due to symmetry. The arrows depict how the E field curls around the flux face of the magnet. When the conductivity is dropped in half to 0.37 S/m, the resulting E field differs from Fig. 1 by a maximum of 0.17%. This was computed by breaking the 1/4 brain region into 778 subvolumes and computing the E field at the center of those volumes.

3.2. Inhomogeneous model 1—concentric spheres

According to Polk and Postow (1996), the conductivity of white and grey matter is 0.48 and 0.7 S/m, respectively. Consider positioning the grey and white matter as a number of concentric spherical bands as shown in Fig. 2. The band pattern was intentionally altered so that the two 1/8 sections would themselves exhibit a contrast.

The surface E field obtained from this model is shown in Fig. 3. Shown in Fig. 4 is a plot of the E field predicted at 3528 points within the volume of this model. The mean absolute value of the difference between these two models is

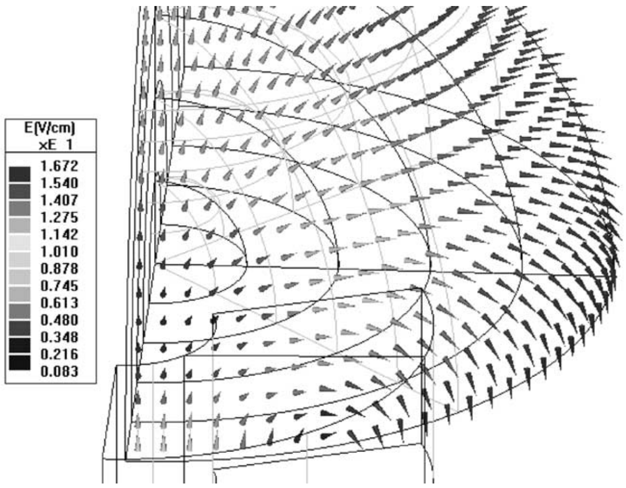

Fig. 3. Surface E field plot for the spherical distribution of white and grey matter.

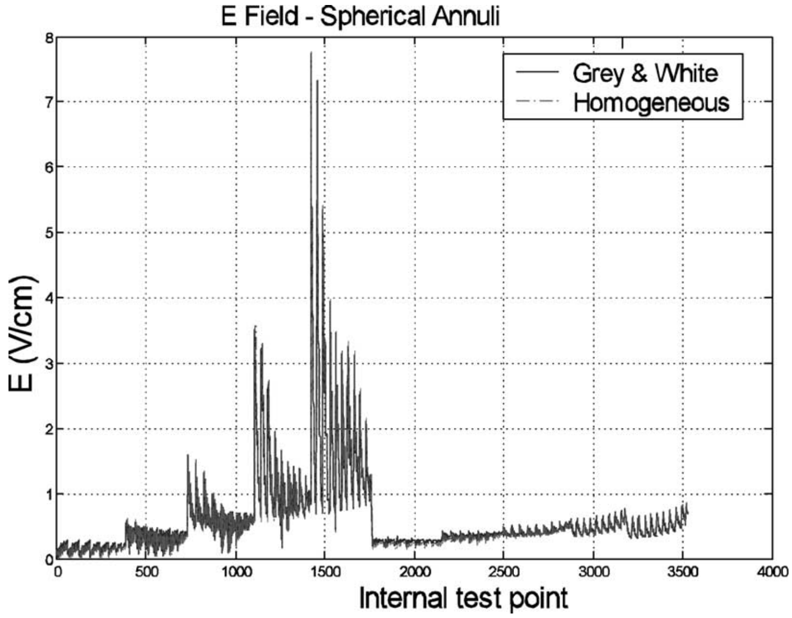

Fig. 4. Comparison of homogeneous and inhomogeneous E fields for a spherical annulus model.

6.8%. Fig. 1 does not account for the differences in grey and white matter, while Fig. 2 does make such a distinction.

Performing a comparison within a volume is challenging, especially if a comprehensive comparison is desired. A simple approach might be to arbitrarily choose a couple of lines within the spherical volume. But the complicated distribution of grey and white matter is critical, allowing for predisposed bias. A better more global approach is obtained by imagining the placement of a test sphere within a cube and then discretizing the cube into 3528 cubic cells, i.e., by breaking each side into 30 increments. After throwing out all the points that fall outside the sphere, 3528 cells remain. The E field is computed at the centroid of those cells. Because of symmetry, the field is plotted only in one quarter of the volume. The plot begins in the first quadrant, and then switches to the second quadrant at point 1655. The steps correspond to different z positions. The larger step cluster of points are each associated with a new setting for z, while the smaller step clusters constitute a new setting for y as suggested in Fig. 5. The numbered arrows correspond to the order of the planes of points so plotted.

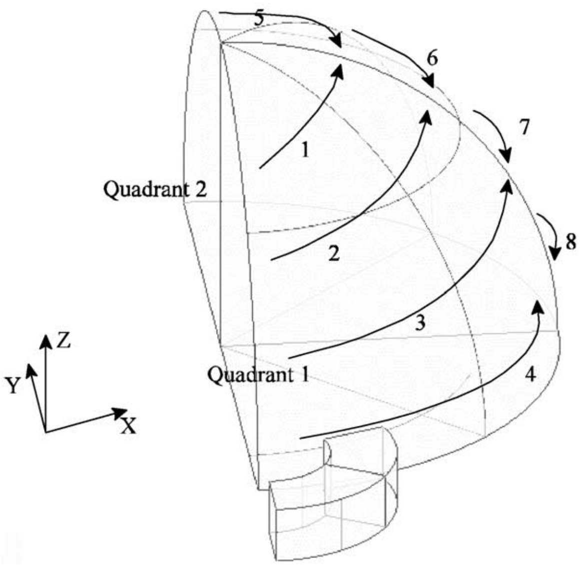

Fig. 5. Order of field point comparison within the volume.

3.3. Inhomogeneous model 2—concentric wedges

Shown in Fig. 6 is yet another model employing a series of spherical wedges. The E field is computed for this model, and compared to that of the homogeneous model in Fig. 7. The mean absolute value of the difference between the two

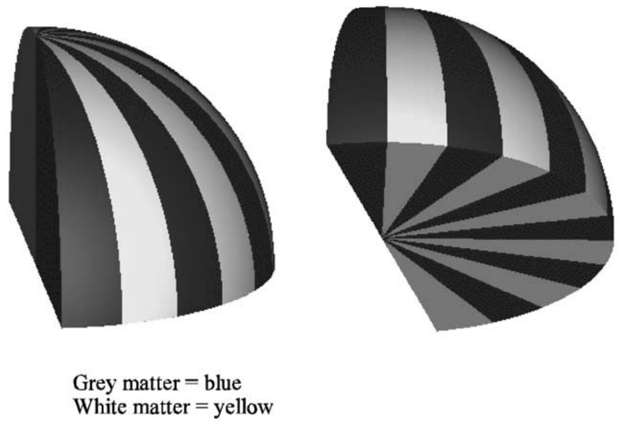

Fig. 6. Brain modeled as a series of spherical wedges, alternating white and grey matter.

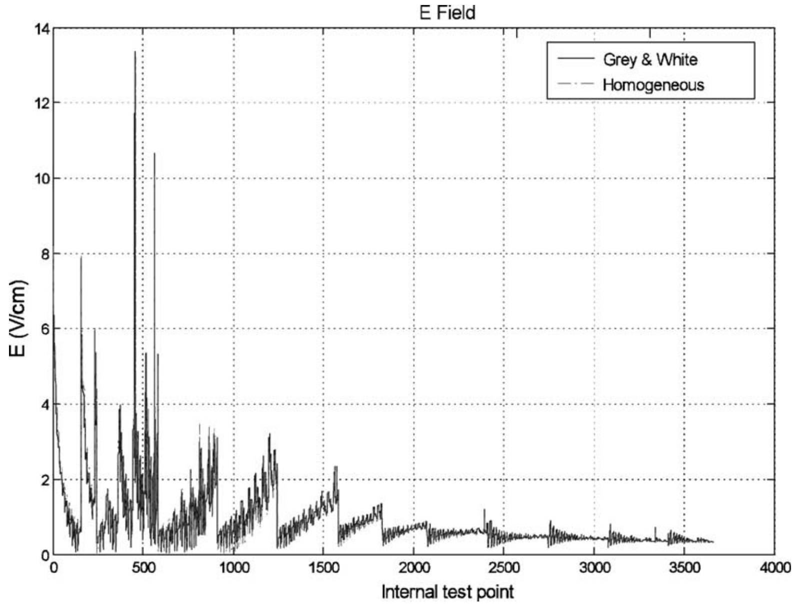

Fig. 7. E field comparison between the wedge shaped model and a homogeneous sphere.

E fields is 7.14%, considering a distribution of 3660 points within the volume.

3.4. Inhomogeneous model 3—concentric wedges with CSF

A final test might suffice to underscore the point being made that paying attention to the distribution of mater in the brain is unwarranted. Baumann et al. (1997) have shown that the electrical conductivity of cerebral spinal fluid is 1.45 S/m at room temperature (25 °C) and 1.79 S/m at body temperature (37 °C) across the frequency range 10–10 kHz. Using the latter value, consider analyzing another wedge shaped model of the brain, this time with CSF distributed in equal volume with white and grey matter as suggested in Fig. 8. In this extreme case, the volume of CSF is assumed to be equal to that of grey and white matter. The difference between the E fields, shown in Fig. 9, increases to a mean absolute value difference of 15.2% due to the higher conductivity of the cerebral spinal fluid. The mean of the difference is only 2.6%. The fact that such an extreme distribution of the three returns such a small difference supports the claim that efforts to accurately model the brain's composition is unwarranted.

3.5. Bounding the maximum electric field variations

When the region of dissimilar conductivity fits totally within another medium, the E field departure from the homogeneous case will be greatest at the point where no tangential electric field exists. The E field will always decrease on the side of the boundary with the smaller conductivity and increase by an equal amount on the side of the larger conductivity, insuring that the normal current density be continuous. For the case of white and grey matter, the maximum departure from the homogeneous case is $0.7/0.48 = 1.458$. So the maximum E field departure will be -22.9% from the homogeneous case on one side and $+22.9\%$ on the opposite side. If a finger like projection of grey matter exists within white matter with a dominant × directed electric field, as suggested in Fig. 10, the change in electric field will be greatest at the ends of the projection. At those points where the induced electric field is tangential to the interface, no change from the homogeneous case will be witnessed.

4. Conclusions

As long as the tissue conductivity differences are small, two homogeneous models will deliver the same induced E

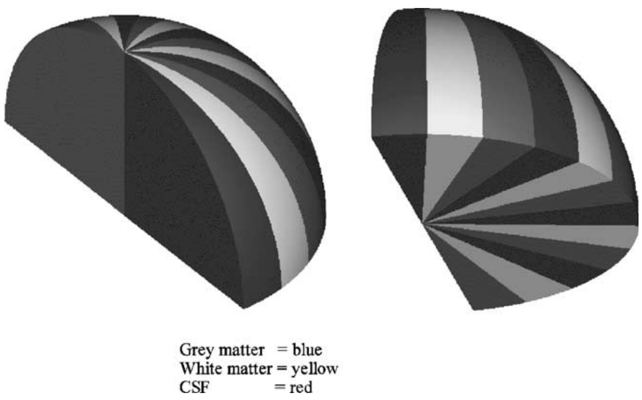

Fig. 8. Combination of white, grey and cerebral spinal fluid.

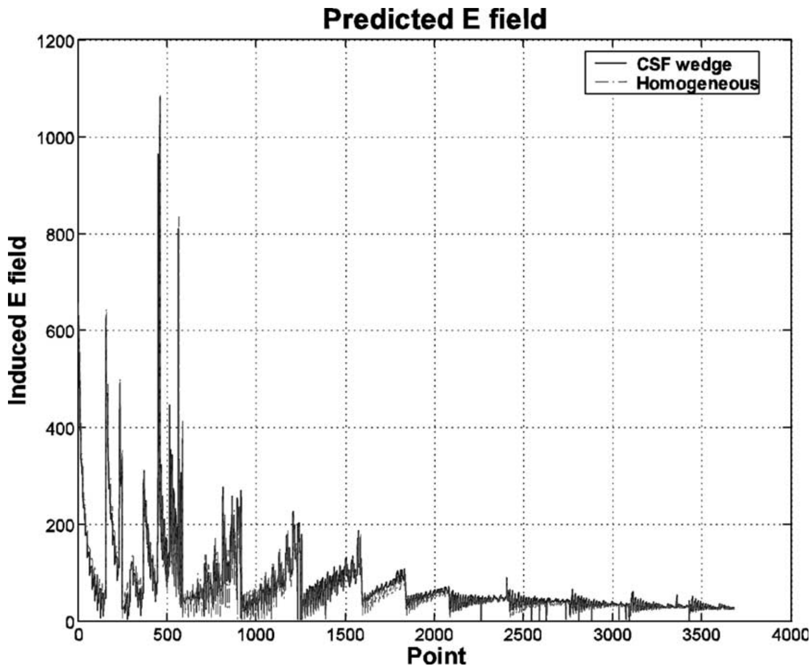

Fig. 9. Induced E field predicted with the wedge model interspliced with CSF.

field regardless of the conductivity distribution. The word small applies when the magnetic field generated by the currents induced is insignificant compared to the stimulation field. The total integrated E field around a loop is fixed by the primary B field. Two models were analyzed containing well defined borders between white and grey matter. Rather extreme distributions of matter in the brain were assumed to determine that the induced fields have a mean variance from

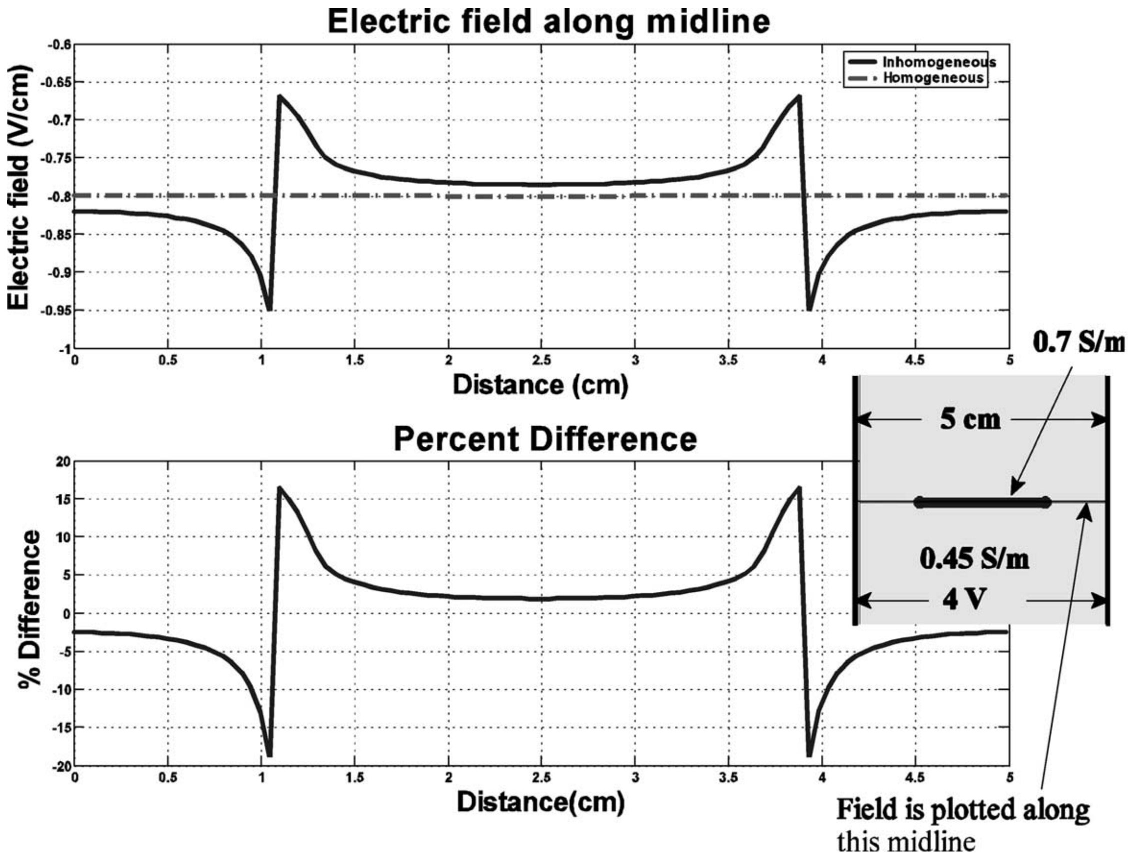

Fig. 10. Electric field change along the midline of a finger like projection of grey matter within white matter.

the homogeneous field of about 7% for white and grey matter, and 15% when cerebral spinal fluid is added. The boundary condition that the normal current density must be continuous dictates the maximum departure from the homogeneous electric field case of 1/2 the ratio of the conductivities of the media involved.

Limitations of these models need to be understood in order to properly interpret these findings. Grey matter has an anisotropic conductivity. Although the calculations were performed using isotropic approximations, the invariance of the induced E field suggests that such differences would not affect the conclusions. The complex patterns of human brain involving gyral folding and layering were not modeled. Additionally, other factors will contribute to whether nerve cells are affected by TMS, including myelination and fiber orientation and morphometry relative to the TMS field (Amassian et al., 1992; Levy et al., 1991).

Acknowledgements

This research was supported by the Defense Advanced Research Projects Agency (DARPA). The numerical calculations were performed using a 3D boundary element solver called Faraday from Integrated Engineering Software, located in Winnipeg, Canada.

References

Amassian VE, Eberle L, Maccabee PJ, Cracco RQ. Modelling magnetic coil excitation of human cerebral cortex with a peripheral nerve immersed in a brain-shaped volume conductor: the significance of fiber bending in excitation. Electroenceph clin Neurophysiol 1992;85:291–301.

Analylist S, Lomarev M, George MS, Bohning DE. TMS-induced versus volitionally-induced BOLD response in the motor cortex: high resolution comparison of location, intensity and repeatability, human brain mapping, submitted for publication.

Baumann SB, Wozny DR, Kelly SK, Meno FM. The electrical conductivity of human cerebrospinal fluid at body temperature. IEEE Trans Biomed Eng 1997;44(3):220–3.

Bohning DE, Epstein CM, Vincent DJ, George MS. Deconvolution of transcranial magnetic stimulation (TMS) maps. Neuroimage 1997;5: S520.

Bohning DE, He L, George MS, Epstein CM. Deconvolution of transcranial magnetic stimulation (TMS) maps. J Neural Transm 2001;108:35–52.

Epstein CM, Davey KR. Iron-core coils for transcranial magnetic stimulation. J Clin Neurophysiol 2002;19(4):376–81.

Levy WJ, Amassian VE, Schmid UD, Jungreis C. Mapping of motor cortex gyral sites non-invasively by transcranial magnetic stimulation in normal subjects and patients. Electroenceph clin Neurophysiol 1991;43: 51–75.

Liu R, Ueno S. Stimulation of the influence of tissue inhomogeneity on nerve excitation elicited by magnetic. Proceedings—20th Annual International Conference—IEEE/EMBS. 1998. p. 2998–3000.

Polk C, Postow E. Biological effects of electromagnetic fields, 2nd ed. Boca Raton, FL: CRC Press; 1996. p. 67.

Roth BJ, Basser PJ. A model of the stimulation of a nerve fiber by electromagnetic induction. IEEE Trans Biomed Eng 1990;37(6): 588–97.

Ueno S, Tashiro T, Harada K. Localized stimulation of neural tissues in the brain by means of a paired configuration of time-varying magnetic fields. J Appl Phys 1988;64(10):5862–4.

Ueno S, Matsuda T, Fujiki M. Functional mapping of the human motor cortex obtained by focal and vectorial magnetic stimulation of the brain. IEEE Trans Magn 1990;26(5):1539–44.

Wassermann EM, Wang B, Zeffiro TA, Sadato N, Pascual-Leone A, Toro C, Hallett M. Locating the motor cortex on the MRI with transcranial magnetic stimulation and PET. Neuroimage 1996;1–9.

Zheng D. Three-dimensional eddy current analysis by the boundary element method. IEEE Trans Magn 1997;33(2):1354–7.