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The Influence of Local Tissue Conductivity Changes on the Magnetoencephalogram and the Electroencephalogram

Der Einfluß der Änderung der lokalen Gewebeleitfähigkeit auf das Elektroenzephalogramm und das Magnetoenzephalogramm

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Key words: Tissue conductivity – finite element method – FEM – EEG – MEG

We examined the influence of local tissue conductivity changes in the vicinity of a dipolar source on the neuromagnetic field and the electric scalp potential using a high resolution finite element method model of the human head. We found that the topology of both the electric scalp potential and the neuromagnetic field (and consequently dipole localization) is influenced significantly by conductivity changes only in voxels adjacent to the source. Conductivity changes in these voxels yield a greater change in the amplitude of the magnetic field (and consequently in the dipole strength) than in the amplitude of the electric potential.

Schlüsselwörter: Elektrische Leitfähigkeit, Finite-Elemente-Methode, FEM, EEG, MEG

Der Einfluß der Änderung der lokalen Gewebeleitfahigkeit in der Nähe einer dipolaren Quelle auf das neuromagnetische Feld und das elektrische Potential wird mit Hilfe eines hochauflosenden Finite-Elemente-Modells des menschlichen Kopfes untersucht. Für Leitfähigkeitsanderungen in den Voxeln, die an der Quelle anliegen, wurde eine signifikante Änderung der Topologie (und demzufolge der Dipollokalisation) des magnetischen Feldes und des elektrischen Potentials gefunden. In diesen Voxeln führt die Leitfähigkeitsänderung zu großeren Änderungen in der Amplitude des magnetischen Feldes (und damit der Dipolstarke) als in der Amplitude des elektrischen Potentials.

1 Introduction

Recently, a study by Baumann et al. [1] found that the electrical conductivity of cerebrospinal fluid (csf) was underestimated by as much as 44 % for nearly two decades. Given the difficulties in estimating in vivo conductivity values [6] it is not unlikely that the conductivity of other tissues is not correctly estimated, either. However, these conductivity values are required for forward and inverse modeling in magnetoencephalography (MEG) and electroencephalography (EEG).

A technique which extracts the conductivity information from diffusion tensor magnetic resonance imaging (MRI) proposed by Tuch et al. [7] is likely to overcome these problems. With this technique, the individual conductivity information of every patient can be used for modeling in MEG and EEG. This technique does not rely on tissue classification or compartment segmentation but on the overlay of anatomical and conductivity information obtained by MRI. Tissue segmentation, therefore, is not needed anymore for those parts of the head where conductivity information is available.

However, this modeling technique is influenced by the signal to noise ratio of the two MRI data sets and the errors introduced by overlaying the anatomical and the conductivity MRI information. In order to quantify the influence of such possible errors we computed neuromagnetic fields and electric scalp potentials by using an isotropic high resolution finite element method (FEM) model of the human head.

Since we expect the strongest influence in the vicinity of the source, the aim of this paper is to examine the influence of local tissue conductivity changes in the vicinity of the source on the neuromagnetic fields and the electric scalp potentials.

2 Methods

The FEM model of the human head was constructed out of 128 sagittal MRI slices with a slice thickness of 2 mm.

Thirteen different tissue types were segmented and assigned to a conductivity value in order to describe the conductivity profile of the human head. The finite element mesh was generated by connecting all slices.

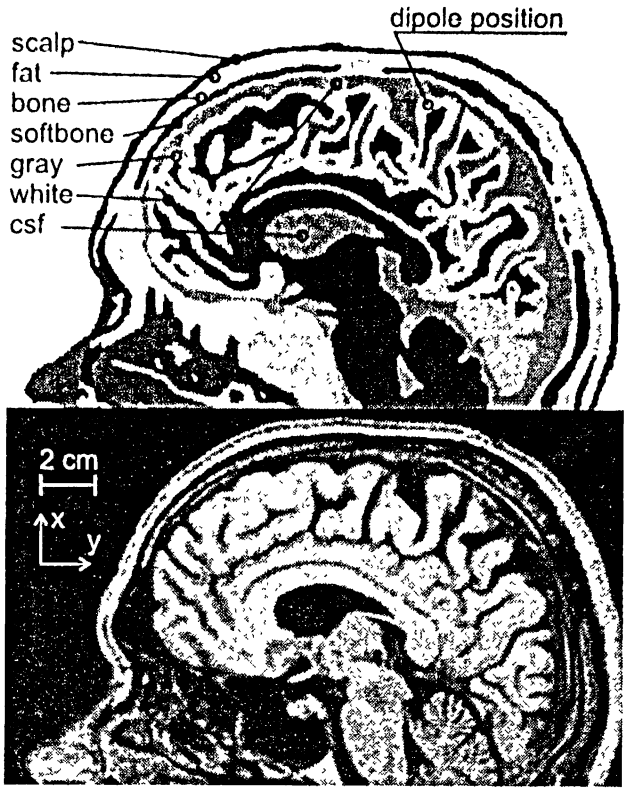

Figure 1. A slice with classified tissues (top) and the corresponding MRI scan (bottom). The original size was 512 × 512 pixel, the bottom portions were cut off. The classified slice represents the FEM model cross section In the classified slice different tissue types are represented by different gray values. The origin of the coordinate system is at the lower left corner of the first slice. The dipole points into the z direction (into the drawing plane)

In this way a uniform grid of 452,162 hexahedral elements (voxels) with a resolution of 2 mm was established. Fig. 1 shows a cross section of the FEM model and the corresponding MRI slice. The model had been used and validated previously [3]. We used a tangential (with respect to the closest point on the scalp) dipolar source in the somatosensory cortex at the position representing the knee. The source was modeled by two fixed voltages at adjacent nodes.

The magnetic fields were computed in a sampling plane (15 × 15 grid, spacing of 1.0 cm × 1.25 cm) located at a distance of 1.2 cm above the head. The sampling plane was brought into such a position that the dipole location corresponded to the middle of the sampling plane. The surface potentials were computed on the scalp close to the position of the dipole (13 × 13 grid, spacing of 0.6 cm, common average reference) .

In order to assess the influence of local conductivity changes, we varied the conductivity value of single voxels in the vicinity of the source. Fig. 2 shows the FEM grid in the vicinity of the source and the voxels in which the conductivity was varied (A–F). The conductivity value of csf (cerebrospinal fluid) was 1.78 S/m, the value of gray matter was 0.33 S/m, and the value of white matter was 0.14 S/m.

The changes in the magnetic field and electric potential were determined by two methods. The changes in field topography were assessed by the correlation coefficient and the changes in field strength were expressed as a percentage of the difference of the absolute amplitudes. In inverse dipole computations changes in the field or potential topography indicate different source positions whereas changes in the field or potential strength indicate differences in the source strength.

All computations were performed at a workstation (IBM RS/6000 model 370). Depending on the initial guess for the voltage distribution and the source model, between 3 000 and 5 000 iterations of the SOR (successive over relaxation) solver were necessary to ensure convergence. The average CPU time for one iteration was 2 seconds.

3 Results

Fig. 3 shows the magnetic field profiles computed with the standard model and the model in which voxel A was changed from gray matter to csf conductivity. The two profiles depicted represent the greatest change observed in this study.

Tables 1 and 2 show the correlation coefficients and amplitude changes for selected conductivity changes of the voxels depicted in figure 1.

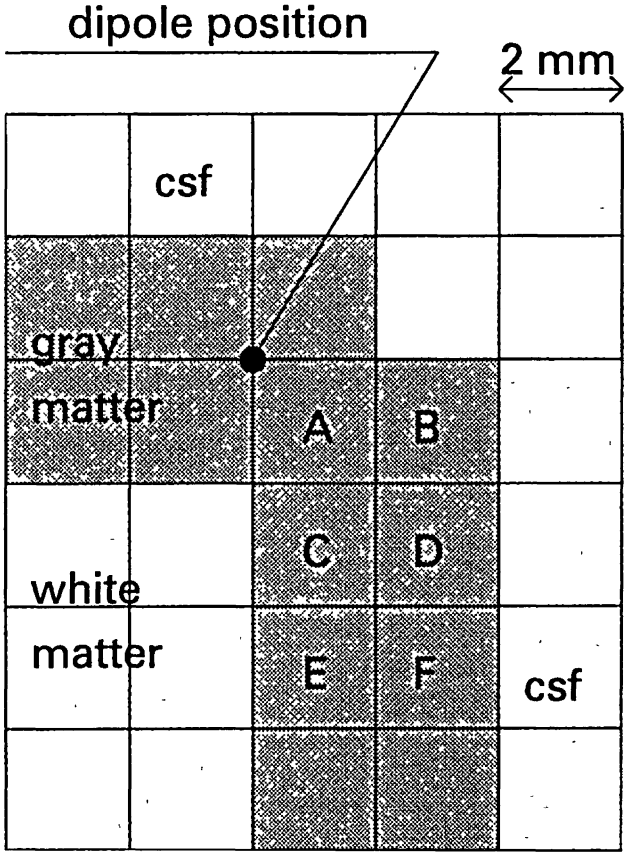

Figure 2. A part of the FEM model of the head in the vicinity of the source (view on the postcentral gyrus in a sagittal slice). The letters A–F identify the voxels varied.

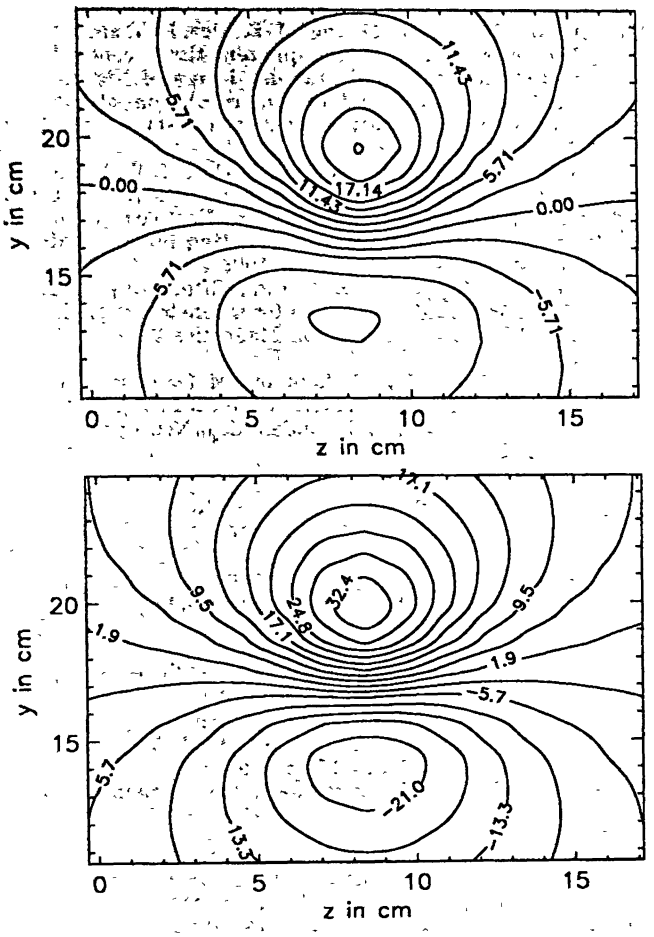

Figure 3. Top: Magnetic field isocontour lines computed with the standard model (mean conductivities). Bottom: Magnetic field isocontour lines computed with voxel A changed from gray matter to csf conductivity. The magnetic field values are in fT.

As expected, a conductivity change in voxel A produced the largest influence on the magnetic field and also on the electric potential. In this voxel, the amplitude of the magnetic field was more strongly influenced than the electric potential amplitude. Furthermore, only conductivity changes in voxel A showed a significant influence on the magnetic field and the electric potential topology.

For a simultaneous change of the voxels B, C, D, E, and F from gray matter conductivity to csf conductivity, the correlation coefficient was above 0.9999 and the amplitude change less than 2 % for both the magnetic field and the electric potential.

4 Discussion

According to our experiences, a correlation coefficient of 0.98 could produce a dipolar source localization error of up to 1.5 cm (average about 5–8 mm) while a correlation coefficient above 0.99 will result in not more than approximately 1 mm localization error [2, 5]. Thus, only changes in the conductivity in the voxels adjacent to the dipolar source will yield errors in source localization procedures. The amplitude changes observed translate with approximately the same percentage value into dipolar source strength changes.

Our results are in good agreement with a recent study investigating the influence of ventricles and lesions on MEG based dipole localization results [9]. They found a significant influence only for such dipoles which are very close to lesion or ventricle. This is also confirmed by earlier work of Ueno et al. [8] showing that even within simple spherical volume conductors inhomogeneities close to sources can significantly affect the measured MEG and EEG.

Previously, we investigated the influence of global conductivity changes on MEG and EEG and found that an accurate modeling of magnetic field and electric potential strength requires accurate knowledge of tissue conductivities, while for source localization procedures this knowledge might not be necessary [4]. In concordance with this previous study the local conductivity changes seem to have a stronger influence on the field strength (i.e. dipole strength estimation) than on the field topology (i.e. dipole localization). In contrast to this previous study, local conductivity changes close to the source have a small but significant influence on source localization, too.

The following limitations of the work presented are important. We considered only a single dipolar source which is the source model most widely used in MEG/EEG studies. The influence of extended or multiple sources remains to be investigated. Additionally, our results are based on isotropic tissue conductivities. We assume that the inclusion of tissue anisotropy will further influence the magnetic fields and electric potentials.

In conclusion, it seems that the relatively small conductivity changes, which could be possibly introduced by low signal to noise ratio or artifacts in the conductivity information obtained from MRI, might not have a great influence on source localizations.

Table 1: Correlation coefficients.

Voxel	Gray matter changed to	Correlation coefficient Electric potential	Magnetic field
A	csf	0.986935	0.989456
A	white	0.999672	0.999418
B	csf	0.999989	0.999991
C	white	1.0	1.0
D	csf	0.999992	0 999990
E	white	1.0	1.0
F	csf	1.0	1.0

Table 2. Amplitude changes.

Voxel	Gray matter changed to	Amplitude change in % Electric potential	Magnetic field
A	csf	10.96	62.62
A	white	2.06	7.57
B	csf	0.02	0.13
C	white	0.05	0.01
D	csf	1.12	0.50
E	white	0.07	0 03
F	csf	0.13	0.05

Acknowledgments

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