

Influence of realistic skull and white matter anisotropy on the inverse problem in EEG/MEG-source localization

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Abstract

The inverse problem in the field of EEG and MEG requires the repeated simulation of the field distribution for a given dipolar source in the human brain using a volume-conduction model of the head. High resolution finite element head modeling allows the inclusion of tissue conductivity inhomogeneities and anisotropies. We will present new approaches for individually determining the direction-dependent conductivities of skull and brain white matter, based on non-invasive multimodal magnetic resonance imaging data, and for generating a high resolution realistically shaped anisotropic finite element model of the human head. Forward calculation and inverse localization errors indicate the necessity of the chosen complex forward model.

1 Introduction

For the EEG/MEG forward problem, the human head has to be modeled as a volume-conductor. The skull is known to have an anisotropic conductivity with a ratio of up to 1:10 (radially:tangentially to the skull surface) [1]. For brain white matter, a similar anisotropy ratio is known, but still, no technique exists for its robust and non-invasive direct measurement. Recently, formalisms have been described for relating the effective electrical conductivity tensor of white matter tissue to the effective water diffusion tensor as measured by Diffusion Tensor Magnetic Resonance Imaging (DT-MRI)[2; 3]. [2] introduced the assumption that the effective electrical conductivity tensor shares the eigenvectors with the effective diffusion tensor of water, which can be measured for white matter tissue by DT-MRI. First studies show that white matter conductivity anisotropy has an influence on the forward solutions for EEG and MEG [4; 5]. In this paper, we will present measurement techniques and methods for obtaining a realistically shaped high resolution volume conductor model of the human head in a non-invasive way with anisotropically conducting compartments skull and white matter. Our goal is the study of the influence of tissue anisotropy on forward and inverse problem in EEG/MEG source localization.

2 Methods

A prerequisite for a realistic modeling of the volume conductor is the segmentation of head tissues with different conductivity properties. Fig. 1 (left) shows an axial slice of the segmented 5-tissue head model from bimodal MRI [5].

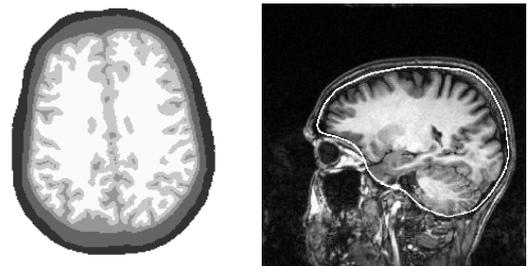


Figure 1: Axial slice of the 5-tissue segmentation result (left). Smooth surface spongiosa model on underlying T1-MRI (right).

2.1 Generation of an anisotropic skull layer

We first describe our modeling approach for the low-conducting anisotropic skull compartment. A first step in the modeling process is the segmentation of inner and outer skull surfaces. We used a bimodal T1-/PD-MRI approach, yielding in particular an improved segmentation of the inner skull surface [5]. We based the determination of skull conductivity tensor eigenvectors on the resulting mesh of a discrete deformable surface model. The deformable model was applied here in order to generate a smooth surface spongiosa model, i.e., a strongly smoothed triangular mesh, which was shrunk from the outer skull mask onto the outer spongiosa surface [5] (Fig. 1, right). For each point in the skull layer, we determine the radial skull direction by means of the normal vector of the surface vertex with minimal distance, and the two tangential directions by vector product. For our simulations, we use conductivity tensors of the form $\sigma = \mathbf{S}\mathbf{S}^T$ with \mathbf{S} the eigenvector matrix as described above and simulated

ratio	Skull		White matter	
	λ_{rad}	λ_{tang}	λ_{trans}	λ_{long}
1:1	0.0042	0.0042	0.14	0.14
1:2	0.0026	0.0053	0.111	0.222
1:5	0.00143	0.0072	0.0818	0.41
1:10	0.000905	0.00905	0.065	0.65

Table 1: Skull and white matter conductivity tensor eigenvalue settings (in $1/\Omega m$). The tensor volume is kept constant for a given anisotropy ratio [5].

eigenvalues $\Lambda = \text{diag}(\lambda_{tang}, \lambda_{tang}, \lambda_{rad})$ (Table 1). λ_{rad} is the conductivity tensor eigenvalue for radial and λ_{tang} for tangential direction.

2.2 Generation of an anisotropic white matter layer

8 DTI measurement sessions, each with 4 axially oriented, $5mm$ thick slices with in-plane resolution of $2 \times 2 mm^2$ were carried out. Coregistered T1 images allowed the registration of the DTI data on the 3D T1 data set. The registered DT data were then resampled to $1 \times 1 \times 1 mm^3$ and each diffusion tensor was rotated with the rotation matrix of the respective registration process via a similarity transform [5]. Fig. 2 (left) shows the trace of the diffusion

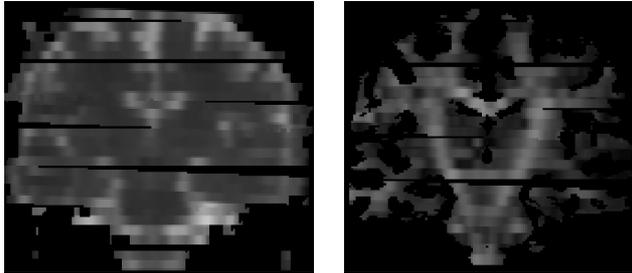


Figure 2: Trace of the diffusion tensor of the 8 registered DT-MRI sessions (left). Fractional anisotropy index of the DT-MRI, masked with the white matter mask of the 5-tissue segmentation result (right).

tensors, i.e., the sum of the diagonal tensor elements, of the 8 registered DTI sessions. Since water diffusion coefficients in CSF are much larger than in the brain, a large contrast is achieved at the brain surface, which allows a quality check of the registration. Fig. 2 (right) shows a map of the fractional anisotropy index ($0 \leq FA \leq 1$) of the registered DT data [5]. The highest FA value was found in the splenium of the corpus callosum, where $FA = 0.74$. For our simulations, we use conductivity tensors of the form $\sigma = \mathbf{S}\Lambda\mathbf{S}^T$ with \mathbf{S} the orthogonal matrix of eigenvectors of the measured diffusion tensors, and simulated eigenvalues $\Lambda = \text{diag}(\lambda_{long}, \lambda_{trans}, \lambda_{trans})$ (Table 1). λ_{long} is the eigenvalue parallel (longitudinal) and λ_{trans} perpendicular (transverse) to the fibre directions.

2.3 FE modeling

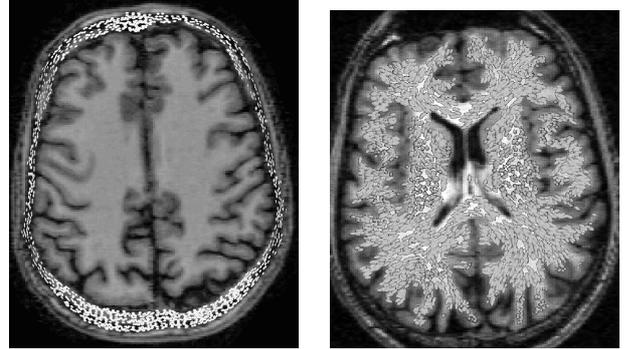


Figure 3: Conductivity tensors in the barycenters of skull elements (left) and white matter elements (right).

We generated a surface-based tetrahedral FE tessellation of the relevant 5 compartments, using [6]. Isotropic conductivity tensors were assigned to skin ($0.33 1/\Omega m$), CSF ($1.79 1/\Omega m$), brain gray matter ($0.33 1/\Omega m$) and ventricular system ($1.79 1/\Omega m$). An anisotropic conductivity tensor was assigned to the barycentre of each finite element in the skull and the white matter compartment (Fig. 3). Tensor validation and visualization was carried out with the SIMBIO visualization module [7]. The resulting model consists of 147287 nodes and 892115 tetrahedra elements. For solving the sparse, large scale, linear FE equation system with many different right-hand-sides, we make use of our parallel NeuroFEM software [7], which is based on a parallel algebraic multigrid solver [8]. The NeuroFEM computation platform used here is an architecturally simple Linux PC-cluster with 100MBit ethernet. For inverse localization, we used a single dipole fit Nelder-Mead simplex algorithm from the SimBio inverse toolbox ST41 [7; 9].

3 Results and discussion

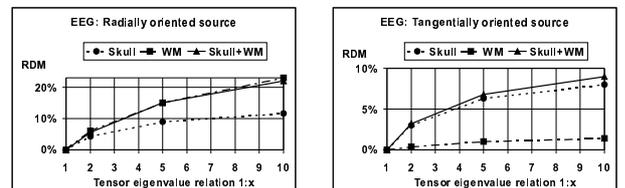


Figure 5: RDM errors in EEG forward result due to anisotropy with ratios from 1:1 to 1:10.

Fig.5 shows the topography error, RDM [5], in 71 electrode EEG forward solutions for a somatosensory source with large radial (left) and large tangential orientation component (right). The errors due to anisotropy effects of skull, white matter (WM) and both skull and WM are presented.

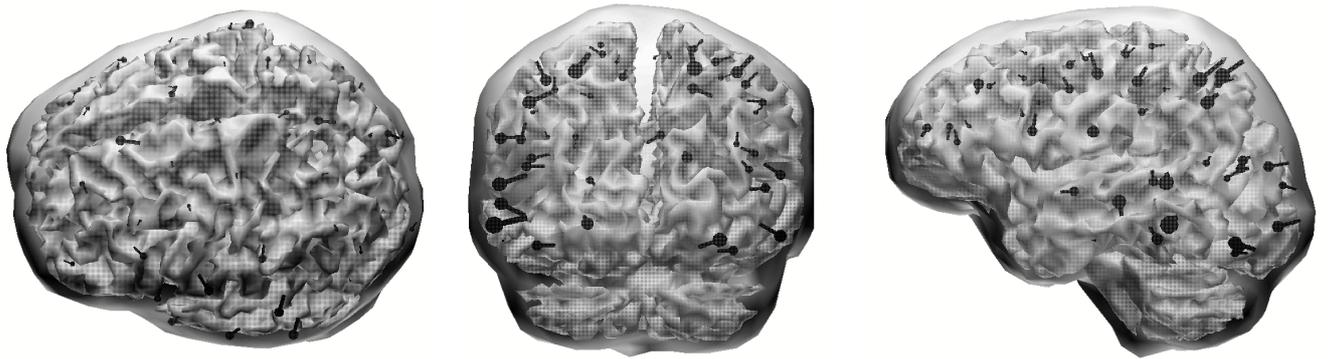


Figure 4: EEG localization errors due to 1:10 anisotropy of skull and WM compartment. The pole is at the position of the simulated dipole, it points to its inverse localization result. Simulated dipoles with large radial (left) and large tangential orientation component (middle and right). Errors are presented on underlying (transparent) WM and inner skull surfaces.

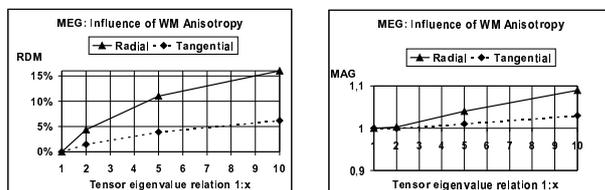


Figure 6: Errors in MEG forward result due to anisotropy with ratios from 1:1 to 1:10.

For the radial orientation, the RDM is mainly due to WM anisotropy, the error of the skull layer was only about half the one of the WM compartment. For the tangential source, the RDM is mainly due to the skull, whereas we found that WM anisotropy is negligible. For both orientations, the magnitude error, MAG [5], is close to the optimum of 1.0 (not shown). An increase of radial or tangential skull conductivity contracts, whereas a decrease spreads out the isopotential distribution on the surface. The isopotential pattern is also distorted, so that an approximation of skull anisotropy effects by means of an increase or a decrease of a scalar isotropic skull conductivity value in boundary element head models seems to be difficult. Fig.6 shows RDM and MAG in whole head BTI 148 channel MEG forward solutions for both sources. With RDM < 1% and MAG \approx 1, skull anisotropy was found to have no influence on the MEG (not shown). For WM anisotropy, the RDM is moderate for the tangential source, whereas it is much larger for the radial one. The larger error can be explained by the fact that tissue anisotropy only influences the secondary (return) currents and that the ratio of the secondary to the whole magnetic flux increases with increasing ratio of the radial dipole orientation component. The MAG is again close to the optimum. Further results concerning the influence of WM and skull anisotropy onto EEG/MEG forward solutions can be studied in [5].

We then computed the EEG at 71 electrodes for 43 neocortical sources with large radial (Fig.4, left) and for 46

with large tangential orientation component (Fig.4, middle and right). For each of them, the EEG in isotropic and in 1:10 anisotropic (both, WM and skull) volume conductors were computed. We validated the single dipole fit method by reconstructing each dipole by means of its EEG result in the isotropic model. Maximal localization error was 0.6mm. We then reconstructed each dipole in the isotropic model by means of its EEG result in the anisotropic model. For the radial sources, the largest localization error is 10.2mm, the average 5.1mm (Fig.4, left), for the tangential it is 17.1mm and 8.8mm, resp. (Fig.4, middle and right). Tangential sources have in particular localization errors in depth. They are localized too deep in the temporal lobe (Fig.4, middle) and too superficial in particular in parietal and occipital areas (Fig.4, right).

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