EEG Analysis on Skull Conductivity Perturbations using Realistic Head Model

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Abstract. Measurement of electroencephalogram (EEG) requires accurate estimation of tissue conductivity. Among the head tissues, skull compartment has less conductivity due to compacta and spongiosa, which impacts on EEG measurement. Therefore, skull conductivity plays a vital role in head modeling, forward computation and source localization. In this study, we have investigated the effects of scalp potentials due to skull conductivity perturbations in realistic head models using different skull to brain and/or scalp conductivity ratio (σ_{ratio}). Several studies used this σ_{ratio} as 1/80, however, other studies found the values of σ_{ratio} between 1/20 and 1/72. Each head model constructed from the values of different σ_{ratio} ranging from 1/20 to 1/72 is compared to the head model constructed from σ_{ratio} = 1/80. The obtained results demonstrated that the skull conductivity perturbations have effects on EEG and the head model constructed from less σ_{ratio} generates larger errors due to higher potential differences.

Keywords: EEG, head modeling, anisotropic conductivity, MRI and FEM

1 Introduction

Tissue conductivity (reciprocal of resistivity) estimation is crucially important in various fields of biomedical research where electroencephalogram (EEG) measurements are involved. Accurate measurement of EEG requires accurate geometry and conductivity distribution [1][2]. Among the head tissue layers, skull shows the lowest conductivity due to its complicated bone structure. The electric potential originated from a current source inside the brain surrounding through the low conductive skull to the higher conductive scalp is known as forward computation [1]-[4]. Therefore, accurate representation and estimations of skull electrical conductivity are essential in developing appropriate EEG forward computation. By inaccurate estimation of skull conductivity resulted errors in EEG [5]-[7]. For

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example, Chen et al. [7] found that inappropriate skull conductivity estimation would cause an average of 6 mm source localization error.

There are several studies on reporting different tissue conductivities. In human conductivity data, there is a significant inter-subject variance [2]. Therefore, the mean values of desired tissue resistivities are used in most of the implementations [5][6][8]. Rush and Driscoll [9] noted that the conductivity ratio (brain : skull : scalp) is $\sigma_{brain} : \sigma_{skull} : \sigma_{scalp} = 1:1/80:1$. Subsequent studies have modified these findings. Hallez et al [5] used σ_{ratio} as 1/16.5 for their source localization on skull conductivity studies. Oostendrop et al. [10] suggested the ratio $\sigma_{brain} : \sigma_{skull} : \sigma_{scalp}$ would be 1: 1/15:1. Baysal and Haueisen [2] found an average *in vivo* skull –brain conductivity ratio of 1:1/23. Lai et al. [11] found the *in vivo* skull conductivity values would be in the range of 1/18 to 1/34. Goncalves et al. [12] proposed that this ratio should be within 1:1/20:1 and 1:1/50:1. In other study [13], they reported 1:1/72:1 ratio using somatosensory evoked fields (SEF) and somatosensory evoked potentials (SEP) data. Studying extensive literature, it is apparent that the estimation of skull conductivity is till on the highest interest for brain science researchers.

The purpose of this study is to examine the EEG by means of forward solution. The main interest is to study how the conductivity ratio of $\sigma_{brain} : \sigma_{skull} : \sigma_{scalp}$ affect the EEG. The organization of this study is as follows. The introduction section describes the necessity of this study with literature review. Methods section describes the realistic head model construction, mesh generation, conductivity estimation, forward modeling and the position of electric source (dipole) in the brain and electrodes (sensors) on head surface. Experiment set up and simulation is described in section 3. Section 4 illustrates the experimental results from our simulation and finally, discussion and conclusion are in section 5.

2 Methods

The reliability of EEG depends on head geometry and accurate estimation of conductivity. To carry out the simulation of skull conductivity perturbations requires the construction of realistic head models, in this case, from magnetic resonance image (MRI) data. MRI is well suited for the segmentation of tissue boundaries.

2.1 Realistic head model construction

To construct a realistic head model requires segmentation of different head tissues. The head tissue segmentation is carried out using the tool BrainSuite2 (http://brainsuite.usc.edu). Firstly, non-brain tissues are removed from the MRI using a combination of anisotropic diffusion filtering, Marr_hildreth egde detection and mathematical morphology [14]. Secondly, each voxel is classified according to its tissue type by combining the partial volume tissue model with a Gibbs spatial prior to produce a classifier, which encourages continuous regions of similar tissue types [14].

Finally, skull and scalp modeling is performed using threshold parameter. Fig 1. shows the segmented brain tissue classification from an MRI data.

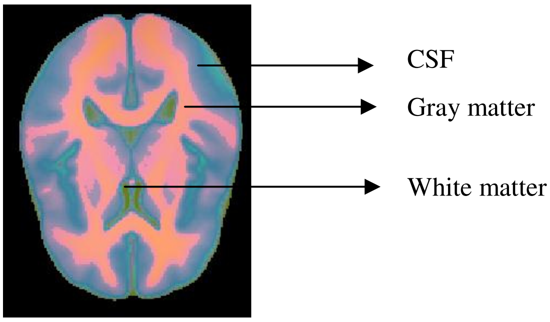


Fig. 1. Different brain component in an MRI image (tissue classification by Brainsuite2)

2.2 Finite element mesh generation

A mesh in this study represents the geometric and electric properties of the head volume conductor. Mesh generation is a prerequisite for finite element (FE) modeling. Mesh generation is performed to create a surface-based tetrahedral tessellation of the segmented head tissues by means of Tetgen software by Hang Si (http://tetgen.berlios.de). The process resulted in a FE mesh with 101K tetrahedra elements from 17K nodes.

2.3 FE conductivity

The tetrahedral or finite elements of head tissues are labeled according to their compartment memberships and later on, the following isotropic conductivities are assigned for reference model based on several literature: brain (σ_{brain}) = 0.33S/m [1], CSF (σ_{CSF}) = 1.0 S/m [1][3][4], skull (σ_{skull}) = 0.0042 S/m (the conductivity ratio for brain : skull : scalp =1:1/80/1) [1][3][4][6], and scalp (σ_{scalp}) = 0.33 S/m [1][3][4][6]. For other models (termed as computed model), σ_{skull} is varied based on different studies while other tissue layer conductivities are constant. In this study, different skull conductivity models are (i) Hallez Model [5] ($\sigma_{brain} : \sigma_{skull} : \sigma_{scalp} = 1:1/16.5:1$), (ii) Oostendrop model [10] (the conductivity ratio is $\sigma_{brain} : \sigma_{skull} : \sigma_{scalp} = 1:1/20:1$), (iii) Baysal model [2] (conductivity ratio is $\sigma_{brain} : \sigma_{skull} : \sigma_{scalp} = 1:1/23:1$), (iv) Lai model [11] with $\sigma_{brain} : \sigma_{skull} : \sigma_{scalp} = 1:1/26:1$ and (v) Goncalves model [12][13] (where the ratio is $\sigma_{brain} : \sigma_{skull} : \sigma_{scalp} = 1:1/47:1$ assuming the proposed mean conductivities).

2.4 FE forward modeling

The standard approach to represent the relationship between electric sources in the brain and bioelectric field based on the quasistatic Maxwell equations is used for the simulation of forward problem [1][6]. In this study, the forward problem is solved by means of quasistatic approximation of Maxwell's equation [1][6]. In this approximation for electric field, the current source density distribution $J^{\Omega}(x_k)$ produces the electric potential distribution $V(x_k)$ in domains $\Omega(x_k)$ is given by Buchner et al. [15] as:

$$\frac{\partial}{\partial(x_k)}\left(\sigma\frac{\partial V}{\partial x_K}\right) = J^{\Omega} \quad \text{................(1)}$$

where σ is conductivity tensor and index k ranges over all spatial dimensions ($x_k = x,y,z$). Dirichlet boundary condition is applied in inner surfaces Γ_I of the boundary $\Gamma(x_k)$ for the specified potentials (α) [15]

$$V\big|_{\Gamma_I} = \alpha(x_k) \quad \text{............................(2)}$$

Neumann boundary condition is applied on the outer surface Γ_O where the mediumis contacted with electrodes and air as [15]:

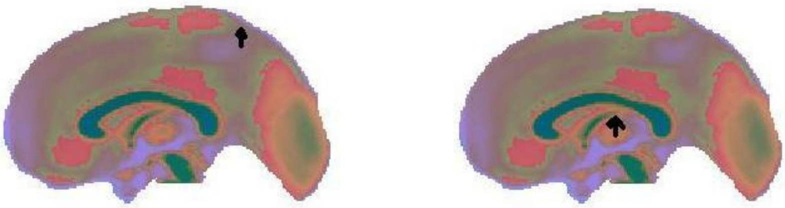
$$\sigma\frac{\partial V}{\partial \mathbf{n}}\bigg|_{\Gamma_O} = 0 \quad \text{........................... (3)}$$

where **n** is the outward unit normal.

For the forward problem, the electric potentials for a volume conductor is computed with known conductivity and current source configuration by solving eqs(1) to (3) by means of an FEM ansatz . A standard variational approach is used to transform the eqs (1) to (3) into an algebraic system of linear equations [3][4][6]. These linear equations are solved by applying the preconditioned conjugate gradient method (pcg) to iteratively solve the linear equations using Cholesky factorization preconditioning with a drop tolerance of $1e^{-4}$.

2.5 Source and sensor positions

The forward simulation is carried out by placing two electric sources (dipoles) in the somatosensory cortex and thalamus inside the brain. All dipoles are unit strength and radially oriented (inferior-superior direction) as shown in Fig. 2. The sensors (electrodes) are logically placed on the scalp according to the international 10-20 system.



(a) (b)

Fig. 2. Electric source (dipole) locations inside the brain: (a) somatosensory cortex and (b) thalamus.

3 Simulation Setup and Experiment

In order to compute the EEG, the procedures illustrated in section 2 are followed. Taking an MRI as an input, segmenting into several head tissue compartments, making mesh generation, assigning conductivity to the individual elements and putting source location inside the brain, we perform forward computation using FEM tool from (http://neuroimage.usc.edu/brainstorm/) . In this study, we consider a single brain compartment in lieu of segmenting gray matter and white. We implement these models using an Intel® dual core 2.0 Ghz processor. A single computation for the FEM modeling requires more than two hours CPU time.

The potentials on scalp are measured by means of 64 electrodes positioned at different places on a head surface. The forward computed data obtained from the reference model and computed models are analysed by calculating relative difference measure (RDM) for the topology error (minimum error: RDM=0) and magnitude difference (MAG) values (minimum error: MAG=1) [1][3][4][6].

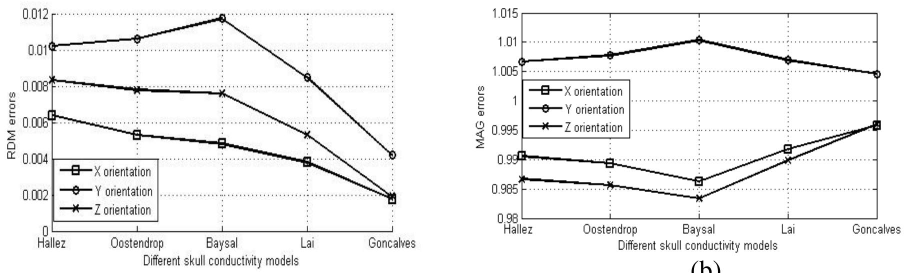
The visualization of the obtained EEGs to observe the differences of scalp potentials produced by different skull conductivity models is also shown in this study. It is performed by adopting and feeding our obtained EEGs to advanced source analysis (ASA) system. ASA is a software package designed for functional brain imaging based on EEG/MEG measurements (www.ant-neuro.com). It represents the axial view (X orientation) of a human head model.

4 Experimental Results

Fig. 3. shows the relative distance measurement (RDM) and magnification (MAG) errors due to skull conductivity perturbations from somatosensory cortex source. The results are shown for the X, Y and Z dipole orientations. Though the errors are less, however, the fewer the σ_{ratio} based conductive models exhibits the fewer the scalp potential differences consequencing fewer errors. For example, RDM errors originated from X oriented dipole generate gradually fewer errors and Hallez model (σ_{ratio} =1/16.5) generates higher RDM errors than all other skull conductivity

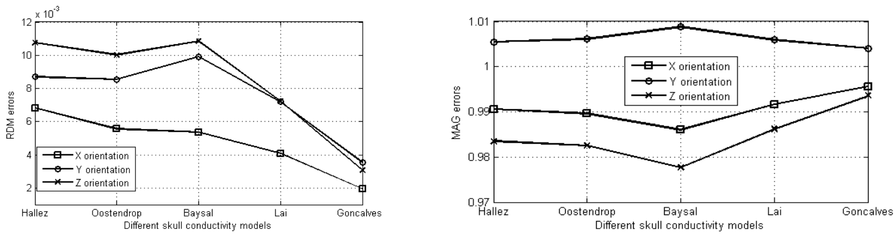
models (one exception for Y directional Baysal Model). For the MAG errors, Hallez model also generates higher errors (away from ideal value 1) except Baysal model with the difference 0.002. Therefore, the closer the σ_{ratio} to reference model generate fewer MAG errors.

The RDM and MAG errors for the thalamus source are shown in Fig. 4. The thalamus source produce similar results like somatosensory cortex source, namely, the fewer the conductive models based on σ_{ratio}, the fewer the errors. By comparing the RDM and MAG errors for both sensors, it is observed that the EEGs obtained from somatosensory cortex source generate fewer errors than those of thalamus sources though the quantity is negligible (0.05%). In this study, we found that EEGs obtained from thalamus sources are more sensible than somatosensory cortex.



(a) (b)

Fig. 3. RDM and MAG errors generated by somatosensory cortex source.



(a) (b)

Fig. 4. RDM and MAG errors generated by thalamus source.

The electric potentials on scalp computed from different conductivity models are shown in Fig. 5. The visualizations are on XY, YZ and ZX dipole planes in the X, Y and Z orientations, respectively.

5 Discussion and Conclusion

This study investigates the effects of skull conductivity perturbations on EEG. The obtained results demonstrate that the RDM errors obtained from Hallez model (σ_{ratio} = 1/16.5) shows higher RDM errors than any other models. It is plausible that computed models would produce some significant errors between them. However, as σ_{ratio} is very close among other models, it generates a very small notable difference. For instance, Oostendrop and Baysal models differ σ_{ratio} = (1/20-1/23 = 0.007). In the conductivity study, Oostendrop model estimates σ_{skull} = 0.0165 S/m while σ_{skull} = 0.0143 S/m is estimated by Baysal model. There may arise a question that why Goncalves model (σ_{ratio} = 1/47) would cause less errors? As this model estimates σ_{skull} = 0.007 S/m, which is very close to reference model (σ_{skull} = 0.0042 S/m), therefore, it produces very close scalp potentials to related models. We can also find it's solution in the visualization of scalp potentials (Fig. 5). We have analyzed and compared the potentials of each electrode for each head model (it is not visualized in Fig. 5) and found that there are some potential changes on several electrodes in different places. The EEG visualization is more informative than RDM and MAG errors for the scalp potentials. The number of skull elements would also be an important factor producing potential differences. The number of skull elements is not so high in this study (5% of the entire head tissues), which may be the other reason for generating close errors for computed models.

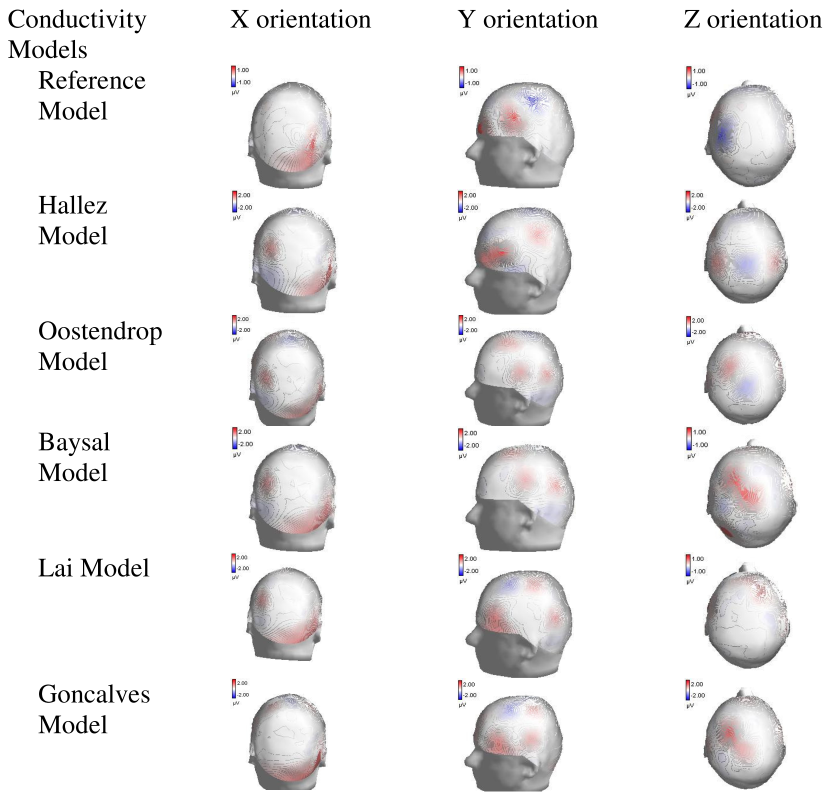


Fig. 5. Visualization of EEG generated by somatosensory cortex from different skull models.

In this study, we have analyzed different EEGs obtained from different skull perturbation conductivity models. The obtained results demonstrate: (i) the skull conductivity perturbations generate variations on scalp EEG and (ii) the closer σ_{ratio} to the reference model, the fewer the errors. In the near future, we shall continue this study on source localization to analyze which skull conductivity model generates more accurate source localization.

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