A Study of White Matter and Skull Inhomogeneous Anisotropic Tissue Conductivities on EEG Forward Head Modeling

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Abstract

The aim of this study is to investigate the effects of white matter (WM) and skull inhomogeneous anisotropic tissue conductivities on human head modeling. The inhomogeneity of WM and skull is included using fractional anisotropy (FA) method and the anisotropy is included according to Volume constraint in the head model construction. A five-layered spherical head model implemented using finite element method (FEM) is used as a volume conductor with a known current source to measure the electroencephalogram (EEG) on the head surface. Statistical measurement techniques are applied to analyze the EEGs obtained from inhomogeneous anisotropic head models and a homogeneous isotropic model. This study finds that the effects of WM and skull inhomogeneous anisotropy on EEG are significant.

***Keywords*:** Fractional anisotropy, finite element method, inhomogeneous anisotropic conductivity, EEG, and forward problem

1. Introduction

The electroencephalogram (EEG) is the measurement of the electric potential differences on scalp caused by neural activity inside the brain. The estimation of these electrical potential differences is known as forward problem [1]. The neural activity is modeled by a distribution of current sources and it's implementation technique is known as inverse problem or source localization [1][2][3]. The inverse problem, solved by iterative solving forward problem, is a valuable tool in pre-surgical evaluation of patients suffering from epilepsy and other neurological disorders [3]. Therefore, the importance of the solution of forward problem is significant. The accuracy of this solution depends on assigning the appropriate conductivity to each tissue which is still a challenging research because of its complicated inhomogeneous and anisotropic properties [1]. However, many head model implementations neglect the inhomogeneous anisotropic conductivities inherent to brain tissues, such as the white matter (WM) and skull. WM has the mean resistivity (reciprocal of conductivity) of 700 Ωcm, with 350 Ωcm and 1050 Ωcm values for lower and upper bounds, respectively, and having the variation of ±50% [4]. Skull resistivity varies between 1360 Ωcm and 21400 Ωcm, with a mean of 7560 Ωcm and a standard deviation of 4230 Ωcm [1]. In WM, the conductivity ratio between longitudinal and transversal directions is 10:1 while it is 1:10 between radial and tangential directions for the skull layer. However, there are no reliable algorithms to measure this ratio non-invasively. Baser *et al* [5] proposed the relation between the electrical conductivity tensors and effective water diffusion tensors measured by diffusion tensor magnetic resonance imaging (DT-MRI). The authors proposed that the tissue conductivity tensors share the same eigen vectors with the effective diffusion tensor. In DT-MRI, the conductivity tensors are not derived directly but inferred from diffusion tensors which describe the movement of water molecules and electrically charged particles (ions) [6]. Later on, Tuch *et al* [7] applied a differential effective medium approach and derived a linear relationship among the eigen values of DT-MRI and the conductivity tensors. However, the eigen values are not equal in the same tissue layer and vary according to

direction. These make the tissue layer as inhomogeneous and anisotropy. The anisotropy depends on the volume of tissues and directions of tissue fibers, which are affected by some conductivity tensor reconstruction obstacles. To overcome these obstacles, C. H. Wolters [2] proposed Volume constraint which restricts the geometric mean with the conductivity tensor.

C. H. Wolters [2] investigated the effects of anisotropic conductivities on EEG for WM and skull tissue layers. The author applied transversal: longitudinal conductivity ratio with the values of 1:1 (isotropic), 1:2, 1:5, and 1:10 for WM tissues. He also applied the same values for radial: tangential conductivity ratio for skull tissues. By applying these conductivity ratios, he found some effects of anisotropy on EEG for Volume constrained WM and skull tissues. Gullmar *et al* [6] also used the same anisotropy ratios for Volume constrained WM on a rabbit head model and found the same results like Wolters. However, they neglected the inhomogeneity of WM anisotropic properties. Li *et al* [8] proposed inhomogeneous anisotropy properties using fractional anisotropy (FA) by step and linear functions for Volume constrained WM. Hallez *et al* [3] implemented five-layered spherical head model to investigate the Volume constrained WM and skull anisotropic tissue conductivities for source localization. They found that neglecting the skull anisotropy resulted on an average of 13.73 mm with a maximum of 24.51 mm dipole localization errors. Most recently, Hallez *et al* [9] implemented a realistic head model incorporating Volume constrained anisotropic WM conductivity and found that WM anisotropy plays a vital role in EEG source localization. Anwander *et a*l [10] investigated the influence of WM and skull anisotropic conductivities and found some significant effects on source localization. In our previous paper [11], we investigated the influence of WM inhomogeneous anisotropic conductivity on EEG using conductivity ratio approximation (CRA) and statistical conductivity approximation (SCA) techniques. We found that there are strong effects of WM inhomogeneous anisotropic conductivity on the scalp EEG. In the paper, we didn't consider skull tissue inhomogeneity and anisotropy. From an extensive literature review, we realize, most of the researchers investigate effects of tissue anisotropy on EEG, however, they didn't mention about the effects of tissue inhomogeneity.

The goal of this study is to evaluate the effects of WM and skull inhomogeneous anisotropic tissue conductivities on the scalp EEG by means of finite element method (FEM). We classify the inhomogeneity by the discrepancy of the ratio longitudinal over transverse eigen values (R_{lt}). We construct a spherical head model by means of varying FA and Volume constraint to simulate the head volume conductor with inhomogeneous anisotropic conductivity settings. We model a current dipole inside gray matter (GM) to simulate brain electrical activity. And we assess the effects of WM and skull inhomogeneous anisotropic tissue conductivities on the scalp EEG by comparing the potential differences generated by the inhomogeneous anisotropic and homogeneous isotropic head models.

The organization of this paper is as follows. Introduction section describes background and the importance of this study with literature review in Section I. Section II illustrates the methods of fractional anisotropy, head model construction, homogeneous and inhomogeneous anisotropic conductivity allocation, forward solution using FEM and similarity measurement techniques. Simulation setup is described in Section III. Experimental results are presented in Section IV. Finally, Section V concludes our research findings.

2. Methods

2.1 Fractional anisotropy (FA)

FA is a technique to measure the extent of the anisotropy property for each voxel (element). Let us suppose that λ_1, λ_2, and λ_3 ($\lambda_1 \geq \lambda_2 \geq \lambda_3$) are the three eigen values of a diffusion tensor matrix and λ is the average eigen value. Then FA is defined as [12]:

$$FA = \frac{\sqrt{3}}{\sqrt{2}} \frac{\sqrt{(\lambda_1 - \lambda)^2 + (\lambda_2 - \lambda)^2 + (\lambda_3 - \lambda)^2}}{\sqrt{\lambda_1^2 + \lambda_2^2 + \lambda_3^2}} \quad \ldots. \quad (1)$$

The FA is in the range of 0 to 1 [12]. A fully anisotropic tissue has a factor FA=1, and an isotropic tissue has a factor FA=0.

2.2 Head model

Anatomically, human head consists scalp, skull, cerebrospinal fluid (CSF), gray matter (GM) and WM. Based on this anatomical concept and different literature [2][3], we consider a five-layered spherical head model. We segment the five-layered sphere and perform mesh generation to produce tetrahedral elements for FEM model. The conductivity tensor in WM or skull of a multi-sphered head model is a diagonal matrix [3]. We assume that the conductivity

tensors share the same eigen vectors with the effective diffusion tensors measured by (DT-MRI) [2][5][6][8]. Then, we consider the conductivity tensor for a WM finite element as [2] [6]:

$$\sigma_{WM} = \mathbf{S}\begin{pmatrix} \sigma_{long} & 0 & 0 \\ 0 & \sigma_{trans} & 0 \\ 0 & 0 & \sigma_{trans} \end{pmatrix}\mathbf{S}^{-1} \qquad \text{……..(2)}$$

where S is orthogonal matrix of unit length eigenvectors of the measured diffusion tensor at the barycentre of the WM finite element, σ_{long} is the parallel (longitudinal) eigen value and σ_{trans} is the perpendicular (transverse) eigen value where $\sigma_{long} >= \sigma_{trans}$. We also consider the conductivity tensor for a skull finite element as [2] [13]:

$$\sigma_{Skull} = \mathbf{S}\begin{pmatrix} \sigma_{tan} & 0 & 0 \\ 0 & \sigma_{tan} & 0 \\ 0 & 0 & \sigma_{rad} \end{pmatrix}\mathbf{S}^{-1} \qquad \text{…….(3)}$$

where σ_{tan} is the tangential (parallel) eigen value and σ_{rad} is the radial (perpendicular) eigen value where $\sigma_{rad} >= \sigma_{tan}$. Assigning the conductivity, we apply FEM for a known current source to solve forward problem.

2.3 Homogeneous anisotropic conductivity allocation

Modeling of the conductivity tensor eigen values is a difficult task involving layer's boundary segmentation, inhomogeneous resistivity, tissue type, and fibre orientation. Therefore, we use isotropic conductivity to simulate the anisotropic conductivity using σ_{trans}: σ_{long} ratios according to literature [2][13] . Then, we calculate the longitudinal and transversal eigen values based on Volume constraint.

2.3.1 Volume constraint

The Volume constraint remains constant of the geometric mean and the volume of the conductivity tensor. The volume constraint is defined as [2][6] ,

$$\frac{4}{3}\pi\sigma_{long}(\sigma_{trans})^2 = \frac{4}{3}\pi\sigma_{iso}^3 \qquad \text{……(4)}$$

where σ_{iso} is isotropic conductivity.

2.4 Inhomogeneous anisotropic conductivity allocation

For the homogeneous anisotropic model, all elements share the same conductivity ratio (R_{lt}) between longitudinal and transversal conductivities. However, R_{lt} varies for inhomogeneous anisotropy. R_{lt} reflects the extent of anisotropy property as FA does, so we set R_{lt} as a variable of FA. We establish the relation between FA and R_{lt} by implementing eq(1). The results are shown in Figure 1. Though the values of FA lie between 0 and 1 [12], we found the values of FA ranging from 0 to 0.9. Considering Figure 1, we define R_{lt} using multi-steps FA function as presented in eq(5).

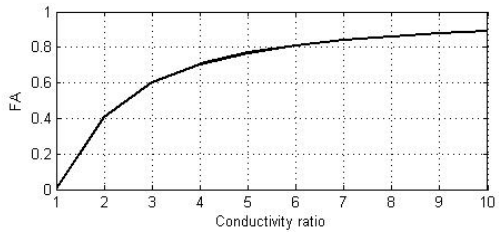

Figure 1: Conductivity ratio Vs fractional anisotropy (FA).

$$R_{lt} = \begin{cases} 10, & FA > 0.8 \\ 6, & FA > 0.7 \\ 4, & FA > 0.4 \\ 2, & otherwise \end{cases} \qquad \text{…….(5)}$$

Using R_{lt}, we generate longitudinal and transversal inhomogeneous conductivities for Volume constrained WM and skull tissue layers.

2.5 Forward solutions using finite element method

The electric potential field generated by a current source in a head is described by the Poission's equation as [2][14]:

$$\nabla.\sigma(\nabla\varphi) = -I_{sv} \text{ in } \Omega \qquad \text{……. (6)}$$

where σ is conductivity tensor, φ is electric potential field, and I_{sv} is internal current source per unit volume due to current dipoles placed within the brain.

The FEM provides a convenient way to assign the conductivity tensor for a small region. Using Galerkin approximation [15] in the framework of FEM and boundary conditions, the solution of eq(6) can be written in a matrix form as [14]:

$$[K]\{\varphi\} = \{F\} \qquad \text{.....(7)}$$

where K is the stiffness matrix [15], φ is the vector of nodal electric potentials, and F is the force vector including the boundary conditions and the current sources. The matrix K contains the head geometry and conductivity information for the forward problem and has the properties being sparse, symmetric, and positive definite. As K is symmetric and positive definite, we use Choleski preconditioner [16] to accelerate the rate of convergence of iterative solvers. We apply the preconditioned conjugate method (pcg) to iteratively solve the linear equation stated in eq(7).

2.6 Similarity measurements

We calculate the electric potential differences produced by a single dipole measured using 64 electrodes positioned at different places on a head surface. The forward computed data obtained from the homogeneous isotropic and inhomogeneous anisotropic model are analyzed by calculating relative difference measure (RDM) for the topology error (minimum error: RDM=0) and magnitude difference (MAG) values (minimum error: MAG=1) of the scalp electric potentials. The RDM and MAG values are calculated as follows [2][6][11][13]

$$RDM = \sqrt{\sum_{i=1}^{n}\left(\frac{ref_i}{\sqrt{\sum_{i=1}^{n} ref_i}} - \frac{meas_i}{\sqrt{\sum_{i=1}^{n} meas_i}}\right)^2} \qquad \text{..... (8)}$$

$$MAG = \sqrt{\frac{\sum_{i=1}^{n} meas_i^2}{\sum_{i=1}^{n} ref_i^2}} \qquad \text{..... (9)}$$

where the values obtained from the homogeneous isotropic model are interpreted as reference (ref) and the values obtained from the inhomogeneous anisotropic model are defined as measurement (meas) [11]. The index i represents the number of electrodes.

3. Simulation setup

First, we implement a five-layered spherical head model according to the radii of Table 1 (row 2) using Matlab [16]. We segment the head model into surfaces, perform tessellation for mesh generation and then, apply a constrained Delaunay tessellation [16] using Tetgen® package provided by BrainStorm [17]. The details of head model construction are described in our previous paper [11]. The mesh generation produces 112 K tetrahedral elements from 19 K nodes as shown in Table 1 (last row). We use these tetrahedral elements for FEM modeling. For homogeneous isotropic model, we assign the mean conductivity to each tissue layer (row 3 in Table 1). However, we assign the conductivities produced by implementing multi-steps FA function (described in eq(5)) to individual elements of WM and skull having other tissue layers isotropic for the inhomogeneous anisotropic head model. We also implement multi-steps FA technique stated in eq(5) using Matlab coded software. After assigning conductivities, we perform EEG forward computation using the adopted FEM tool provided by BrainStorm [17] for a fixed current dipole inside the GM. We assume that the dipole is located in axial, coronal and sagittal planes with the azimuth angle of π/4 and elevation angle of π/5, and having the unit magnitude. A dipole can be decomposed into three orthogonal dipoles along the main axes [9]. For each dipole location, we consider three orthogonal orientations: along left-right (X orientation), along back-front (Y orientation) and along bottom-top (Z orientation). Figure 2 shows the logical head model with different tissue compartments and dipole orientations. We measure the EEGs by putting 64 electrodes on head surface. Finally, we apply RDM and MAG techniques to analyze the results. We perform this computation using an Intel® dual core 2.0 Ghz processor with 2GB RAM. It takes approximately an hour to execute each computation.

Table1: Head model parameters

	Scalp	Skull	CSF	GM	WM
Radii (cm)	9.2	8.4	8.0	7.6	5.0
Means (S/m)	0.33	0.0042	1.0	0.14	0.33
Elements	19397	24563	21379	20674	26841

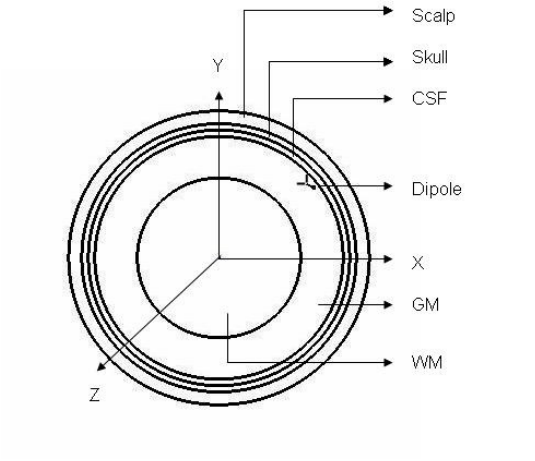

Figure 2: Head model construction showing different head compartments and dipole (source) location.

4. Experimental results

To study the influences of inhomogeneous anisotropic WM and skull tissue conductivities, we carry out four independent experiments.

1. We measure EEG from the reference model.
2. We measure EEG from the measurement model where inhomogeneous anisotropic conductivity is assigned to WM while other tissue layers are homogeneous and isotropic.
3. We measure EEG by assigning inhomogeneous anisotropic conductivity to skull while other layers are homogeneous and isotropic.
4. We measure EEG by assigning the WM and skull inhomogeneous anisotropic conductivities considering other layers homogeneous and isotropic.

Table 2 shows the RDM and MAG errors caused by the inhomogeneous anisotropic WM conductivity generated using the Volume constraint. We found that the RDM (1.59% ~ 18.87%) and MAG (0.95 ~ 1.12) values are far from their ideal values, 0 and 1, respectively. These results indicate that WM inhomogeneous anisotropy affects the scalp EEG strongly. The longitudinal inhomogeneous conductivity produces fewer errors than those of transversal conductivity. We found that the obtained results for WM inhomogeneous anisotropic models are consistent with those published from different literature for WM anisotropic model neglecting inhomogeneity [2][10][13]. We conclude that WM inhomogeneous anisotropic conductivity has significant effects on EEG and WM transversal inhomogeneous conductivity has more effects on EEG than longitudinal inhomogeneous conductivity.

Table 2: RDM and MAG values generated by inhomogeneous anisotropic WM conductivity

Conductivity	Dipole orientation	RDM	MAG
Longitudinal	X	4.04%	1.02
	Y	5.91%	1.12
	Z	4.21%	1.03
Transversal	X	18.87%	1.07
	Y	1.59%	0.97
	Z	7.3%	0.95

Table 3 shows the topological and magnitude difference errors due to anisotropic inhomogeneous skull conductivity. Note that the RDM values range from 4.37% to 17.19% and MAG values are between 0.84 and 1.11. It demonstrates that the effects of inhomogeneous anisotropic skull tissue conductivity on EEG are significant. Radial inhomogeneous conductivity results more errors than tangential inhomogeneous conductivity. These results are also consistent with different literature where inhomogeneity was absent [2] [13].

Table 3: RDM and MAG values generated by inhomogeneous anisotropic skull conductivity

Conductivity	Dipole orientation	RDM	MAG
Radial	X	17.19%	0.89
	Y	7.93%	0.84
	Z	7.17%	0.96
Tangential	X	8.18%	1.09
	Y	4.37%	1.11
	Z	4.64%	1.03

Table 4 shows the RDM and MAG errors generated by combining the inhomogeneous anisotropic WM and skull tissue conductivities. The longitudinal conductivity produces 1.23% to 5.9% RDM and 0.95 to 1.01 MAG errors while the transverse conductivity produces 7.03% to 20.39% RDM and 1.04 to 1.09 MAG errors. It clearly demonstrates that the combination of inhomogeneous anisotropic WM and skull conductivities have significant effects on EEG. By contrasting Table 2 and Table 4, we found the similar results, namely, EEGs are more affected by transversal conductivity than longitudinal conductivity.

Table 4: RDM and MAG values generated by inhomogeneous anisotropic WM and skull conductivities.

Conductivity	Dipole orientation	RDM	MAG
Longitudinal	X	5.9%	0.95
	Y	1.23%	1.01
	Z	4.48%	0.99
Transversal	X	20.39%	1.09
	Y	8.96%	1.08
	Z	7.03%	1.04

5. Conclusion

In the present study, we have investigated the influences of WM and skull inhomogeneous anisotropic tissue conductivities on EEG forward computing using a spherical head model by means of Volume constraint. We have implemented the multi-steps FA to generate Volume constrained anisotropic conductivity and various anisotropy ratios to generate inhomogeneity. From our experiment results, we find that WM inhomogeneous anisotropic tissue conductivity results maximum 18.87% RDM and 1.12 MAG errors and skull inhomogeneous anisotropic tissue conductivity results maximum 17.17% RDM and 1.11 MAG errors. Combining WM and skull inhomogeneous anisotropic conductivities result maximum 20.39% RDM and 1.09 MAG errors. This study also finds that the WM inhomogeneous transversal and skull inhomogeneous radial conductivities have more effect on EEG.

6. Acknowledgement

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