EEG human head modelling based on heterogeneous tissue conductivity

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Abstract

This paper studies the effect of heterogeneous tissue conductivity in a human head model for the EEG forward problem. Firstly, the tissue heterogeneity in conductivity was characterised from measured data in the literature. Then a method was developed to include this feature in modelling. Finally, the effect of tissue heterogeneity on EEG signals was studied. Based on these studies the paper concludes that the inclusion of tissue heterogeneity is significant in accurate head modelling for the EEG problem.

Key words EEG, human head model, tissue conductivity, realistic head model

Introduction

The activities of the brain, either due to background activity or as a response to some stimulus, cause the electric potentials of the cells within that region to change over time. These electrical activities in turn produce an electric field that affects (to a small degree) the entire human head. The electrical potentials within this field conduct outwards through brain tissue, enter the membranes surrounding the brain, and continue on through the skull to appear at the scalp. At this point they are reduced from the milli-volt range (of the membrane potential and action potential) to a few micro volts. Although exceptionally small in magnitude, this field can be detected by applying electrodes to the scalp, and recording the electric potential at each electrode. This recording is known as an electroencephalogram or EEG, and gives a picture of the activity of the brain over time.

The extremely small potentials recorded on the scalp are often outweighed by the electric fields produced by unwanted activity (eye blinking, muscle movement etc.). This means that the signal-to-noise ratio of a typical EEG recording is very small, which makes it extremely difficult to analyse the background brain activities. This study is part of the work to develop mathematical descriptions of the origins of bioelectric signals and provide researchers in bioengineering and medicine with new diagnostic tools. It focuses on the modelling and simulation of macroscopic bioelectric fields in the human EEG. This paper reports the attempts to develop new approaches to model the brain's electrical activities more accurately by addressing heterogeneous tissue conductivity in the head.

In the field of EEG, computational models of bioelectric phenomena from sources in the brain have existed for decades. The size and scope of these models have been limited by contemporary computational resources and by the numerical algorithms utilized to approximate the continuous field equations. It has been shown that the electric signals in the brain, when viewed macroscopically, can be described as a solution to a quasistatic Poisson's equation [1-2].

Poisson's equation, a mathematical description of a typical bioelectric volume conductor problem, can be written as

$$\nabla \bullet \sigma \nabla \Phi = -I_v \qquad \text{in } \Omega \text{ domain} \qquad (1)$$

where Φ is the electrostatic potential, σ is the electric conductivity tensor, and I_v is the current per unit volume defined within the solution domain Ω. The problem statement is to solve equation (1) for Φ with a known description of I_v, σ and the Neumann boundary condition:

$$\sigma \nabla \Phi \bullet \vec{n} = 0 \qquad \text{on} \qquad \Gamma_T \text{ surface} \qquad (2)$$

which says that the normal component of the electric field is zero on the surface interfacing with air (here denoted by Γ_T).

While the analytic solutions of such elliptic partial differential equations are not difficult to achieve for simple

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geometries such as spheres and cylinders, difficulties arise when considering the complex geometries associated with physiological structures. Realistic geometry poses a significant challenge to researchers trying to accurately approximate the bioelectric fields within them. Even more challenges are posed when realistic conductivities are included in the bioelectric field problem. Such challenges include the construction of anatomically accurate model geometries of each compartment of the head; the specification of the material properties, most of which are heterogeneous and some of which are anisotropic; the numerical approximation of the biophysical field equations; and the large-scale nature of the computations.

Head shape can be obtained from magnetic resonance images (MRI) by extracting surface boundaries for the major tissues, such as scalp, skull, CSF and brain. However, most tissues in the brain are neither homogeneous nor isotropic. To account for tissue heterogeneity, more tissues types than the four common ones (scalp, skull, CSF and brain), as well as tissue subtypes such as hard or soft bones, muscles, blood vessels etc., need to be extracted from MRI scans. Some recent studies have identified as many as 12 tissues, plus air in the sinuses[3]. Furthermore, some tissues, such as white matter in the brain, muscles in the scalp and even the bones in the skull, are known to be anisotropic. Thus, even within the same tissue, the conductivity varies with density, microstructure and orientation. The difficulty in creating an accurate and realistic head model is that the conductivities of the various tissues in the head cannot be measured in the living patient or subject[4]. A full accounting of the anisotropy and the heterogeneity for all the tissues in the human head has yet to be performed.

In this paper, this Introduction section provides a brief on the problem of modelling bioelectrical phenomena in the human head. The next section introduces the tissue conductivity, and the pseudo-conductivity method to include these heterogeneous tissue conductivities. Then, a series of head models which range geometrically from the three-sphere model to the realistic model (that is, realistic in shape) are developed in the following sections. The influences of tissue heterogeneity on EEG signals are investigated in simulations. The last section provides the main conclusions based on the results of these investigations.

Heterogeneous tissue conductivity

The finite element method (FEM) is a powerful numerical technique for obtaining approximate solutions to boundary value problems of mathematical physics. This method has a history of about 40 years. The main strength of the FEM is that it computes an estimate of the potential field around each element based on the material properties of that individual element. This property is helpful in that it allows a different conductivity tensor to be allocated to each element within the model.

To apply FEM to the forward computation of EEG, the head should be modelled as a large number of elements with each of these elements representing a different portion of the head with its own conductivity. Not only do the elements representing different tissues have different conductivities, but so do the elements representing the same type of tissues. The latter is due to the complex heterogeneity of the tissues. For instance, the elements in the brain may have different conductivities, since they may contain different proportions of blood vessels, white matter, grey matter, etc.

Detailed measurements on specific regions of brain tissues, particularly white matter, grey matter, dura and pia mater, demonstrate the local variations which may occur over small distances in a tissue which may well be otherwise considered as having constant conductivity[5-6]. Experimentally measured values of conductivity for grey matter increase as a function of the measuring signal frequency (e.g., 0.33 S/m at 5 Hz, 0.43 S/m at 5 kHz, etc.). White matter has a conductivity of 0.76 S/m at 5 Hz, and has been shown to be anisotropic with the ratio of conductivities varying between 5.7 S/m and 9.4 S/m[7]. The conductivity of the cerebrospinal fluid (CSF) surrounding the brain is generally accepted to be 1.0 S/m [8-12]. However, the conductivity values of CSF measured in different cases varied around 1.0 S/m. The variation is typically 10% and sometimes up to 20%. In the case of the skull, the element conductivity may differ for elements composed purely of cancellous bone or compact bone, or some combination of the two. Its resistivity (the inverse of conductivity) varies between 1360 Ωcm and 21400 Ωcm, with a mean of 7560 Ωcm and a standard deviation of 4230 Ωcm [13]. All models reported in the literature use the value of 0.33 S/m for the scalp conductivity. No allowance has been made for the conductivity of the underlying muscle (0.0076 to 0.52 S/m), or subcutaneous fat (0.02 to 0.07 S/m) [7].

The conductivity differences of tissues can be modelled by different compartments in the model, as different tissues can be identified from medical image data. How can the conductivity differences of different elements in the same tissue be represented, given that the current medical image data cannot tell us the differences between the elements in structure, composition and sometimes orientation? Given that the conductivities of the elements for the same tissue are different but relatively close in value[5-7, 13], the conductivities of the elements in a tissue can be assumed to follow a distribution whose density can be described as $f(x,\mu,\sigma)$ [14], where x is a random variable representing the conductivity, μ is the mean of x, and σ^2 is the standard deviation of x. Changing μ corresponds to increasing or decreasing the average conductivity. For small σ^2, the conductivities of the elements within a tissue are tightly centred around the mean, and for $\sigma^2=0$, conductivities everywhere within that tissue are the same – as assumed in the current literature. Models based on the premise that $\sigma^2=0$ are referred to as piecewise homogeneous conductivity models. Conversely, with increasing σ^2, the conductivities of the elements are more widely distributed.

From the above assumptions, a set of statistical

parameters (namely μ and σ^2) for a tissue type can be estimated from the limited measured data available for that tissue in the literatures. A range of conductivities - namely the pseudo-conductivities - can then be generated to fit the distribution which is specifically defined by μ and σ^2, for example, the proposed Gaussian distribution[14]. These pseudo-conductivities are allocated to the component elements belonging to that tissue. Thus a model with tissue heterogeneity, here referred as a heterogeneous conductivity model, can be developed. The keys to the method are the distribution and its parameters μ and σ^2. Once the distribution function and the parameters are given, all the properties of the pseudo-conductivity can be determined. The parameter μ represents the average conductivity of the tissue, and its influence on the computed potentials has been extensively studied previously through many piecewise homogeneous models[16-20]. Therefore, the current study focuses on the parameter σ^2 to explore how the spread of the conductivities affects the computed solution.

Models with a single homogeneous/heterogeneous tissue

The aim of this section is to test the effects of each tissue's heterogeneity on computed electrical potential fields associated with the EEG forward computation. Although it is currently not possible to measure scalp potentials and the dipoles in the brain simultaneously, and thus data for absolute validation are not available, what the results can establish is the relative effect of selectively including or excluding a wide variety of heterogeneities in a detailed model of the human head.

Modelling

A three-layer sphere head model was generated using the commercial package ANSYS. The tetrahedral element was used in the mesh generation. Each layer was meshed separately, so that the scalp, skull and brain are composed of 7534, 9188 and 10756 tetrahedral elements respectively, a total of 27478 elements. Since the concentric sphere model is symmetric, the finite element model needs only to include one half of the symmetric domain. The simulations were conducted within the upper half of the concentric sphere model. An analysis program was developed based on the algorithm provided by Yan *et al*, where the electrical source (current dipole) is modelled as a Delta function[11]. For the linear base function used in this algorithm, the derivatives are constant. This saves a lot of computation. The program was implemented using Matlab on a Unix workstation. The parameters of the model and the number of elements are listed in table 1.

Simulations

To study the effects of each tissue's heterogeneity on the electrical field in the head, two types of simulation were carried out. First, a piecewise homogeneous model was used, then a single heterogeneous layer was included for comparison. "Including an heterogeneous layer" actually means assigning a set of pseudo-conductivities to all the elements in a layer in the head model. The second type of simulation is the complement of the first. Its goal is to evaluate the effect of removing a single heterogeneity from an otherwise completely heterogeneous model. In this case, the reference model includes all available heterogeneities.

Table 1. *Three-layer sphere model parameters.*

Layers	Brain	Skull	Scalp
Radii(cm)	8.7	9.2	10
Means(S/m)	0.33	0.0042	0.33
Variances	0.0099	4.478×10^{-6}	0.0099
Elements	10756	9188	7534

The results evaluated from the first set of simulations demonstrated the effect of adding heterogeneity to the model, which are listed in table 2 as the root mean square difference (D_{rms}) and the relative difference (D_{rel}), which are defined by Yan *et al* as the follows[11].

$$D_{rms} = \sqrt{\frac{\sum_{i=1}^{N}(\Phi_i^{ref} - \Phi_i)^2}{N}} \tag{3}$$

$$D_{rel} = \sqrt{\frac{\sum_{i=1}^{N}(\Phi_i^{ref} - \Phi_i)^2}{\sum_{i=1}^{N}(\Phi_i)^2}}, \tag{4}$$

where Φ_i^{ref} is the potential obtained from the reference model, and Φ_i is the potential from the model being compared to the reference. The comparison is based on the simulation results generated from each of the three spherical models.

In the second simulation, the selective removal of heterogeneity reinforced some findings observed when individual heterogeneity was added. The results are also listed in table 2.

Table 2. *Result evaluation in D_{rms} and D_{rel}.*

	Single heterogeneous vs Homogeneous			Single homogeneous vs Heterogeneous		
	Brain	Skull	Scalp	Brain	Skull	Scalp
D_{rms}	0.0168	0.0327	0.0201	0.0275	0.0532	0.0333
D_{rel}	28%	61%	36%	31%	58%	41%

Models including only the heterogeneity of different tissues always incur significant differences compared to those that incorporated a more complete representation of head heterogeneity. The comparisons made among the models with different conductivity combinations demonstrated that the effect of various heterogeneities in head tissues is very strong. Thus, most of the current models which neglect this effect are inaccurate**.** The results

show that the skull has the largest impact on the head potential distribution. Given the same strength of heterogeneity, the outer layer, viz the scalp, affects the computational potentials more than the inner layer, viz the brain. It also shows that the size of the effect is not negligible in all tissues. This study suggests that the accurate representation of the heterogeneity of each tissue has a significant effect on the accuracy of the EEG forward computation.

Models with realistic geometry

Most realistic bioelectric volume conductor models are based upon MRI or CT images of patient anatomy. The realistic head model employed here is from the MRI slices of a woman's head. There are 20000 triangular elements which connect together to form the head volume. The original data was obtained as an ASCII text file from The Neurosciences Institute, San Diego, California, U.S.A. The rearrangement and refinement were completed with the help from University of Surrey, Guildford, Surrey, UK.

Modelling

An essential goal of this section is to apply the FEM to a volume conductor which closely approximates the actual shape of the head. The first problem was approached by finding and connecting the triangles on each tissue surface. For example, the triangle would be classified as skull surface if its three nodes belong to the skull. The properties of each node were given in the original data file. Fortunately, all the triangles are on the surfaces, and there is no case where three nodes of a triangle belong to different tissues.

Having the surface represented, the next step is to construct a coarse mesh of tetrahedra from the boundary points and then determine tetrahedra within the surface of interest. These interior tertrahedra are used in subsequent steps, which iterate over the generation of a new point and subsequent tessellation until certain spacing criteria are satisfied. The spacing criteria are what ultimately determine the size of the mesh.

Our final model has been built up with 2254 nodes, and has 10640 tetrahedral elements. In the scalp and skull volume, each element has a triangular face on one volume boundary and the remaining apex on the opposite boundary, thus leading to 612 nodes on each surface. The remaining 418 nodes are spread in the brain volume with the tetrahedral elements getting larger toward the centre of the model. There are 3153, 3153 and 4334 elements in the scalp, skull, and brain volumes, respectively.

To investigate how tissue heterogeneity affects the computed potentials in a realistic head geometry, we assigned homogeneous and heterogeneous conductivities to the elements of the scalp, skull and brain tissues to form a piecewise homogeneous model and a fully heterogeneous model respectively.

The comparison is carried out between the results of the model with piecewise homogeneous conductivity and the models with different pseudo-conductivity using the criteria RDM and MAG. The two criteria were introduced by Meijs *et al* in 1987 [15] and used by Yan *et al* in 1991 [11]. The definitions are given in the following equations.

$$RDM = \sqrt{\int_{\Omega}\left(\frac{\Phi_r}{\sqrt{\int_{\Omega}\Phi_r^2}} - \frac{\Phi_p}{\sqrt{\int_{\Omega}\Phi_p^2}}\right)^2} \tag{5}$$

$$MAG = \sqrt{\frac{\int_{\Omega}\Phi_p^2}{\int_{\Omega}\Phi_r^2}} \tag{6}$$

where Φ_r is the potential from the reference model - piecewise homogeneous head model, and Φ_p is the potential from the model with pseudo-conductivity.

The RDM quantifies the errors or differences in topography, whereas the MAG represents the magnification factor of the pseudo-conductivity model solution Φ_p with respect to the reference model solution Φ_r. Ideal values for RDM and MAG are 0 and 1, respectively.

The homogeneous model was formed by assigning all the elements with the mean conductivity of the appropriate tissue. The heterogeneous model was formed by assigning each element a pseudo-conductivity which was created using the parameters listed in table 3.

Table 3. *Realistic head model parameters.*

	Scalp	Skull	Brain
Mean (S/m)	0.33	0.0042	0.33
Standard deviation	0.099	0.0021	0.099

After the homogeneous and heterogeneous models were formed, the source, a radial dipole, was placed at various eccentricities from 0 to 0.85 to form our EEG forward problem. For each model and dipole a set of potential values was computed. The results have been compared in terms of RDM and MAG.

Simulation

Figure 1 illustrates the RDM differences found on the scalp surface for the solution. Figure 2 shows the MAG magnification factor of the solution on the scalp surface.

Figures 1 and 2 show similar results, that is the more eccentric the dipole, the greater the difference. RDM differences lower than 5% were obtained for eccentricities less than 0.6. However, RDM begins to increase dramatically from the point where eccentricity equals 0.4. The general tendency of RDM differences is to rise as an exponential function. It reaches 21% at the point where eccentricity equals 0.85, i.e. the point at which the dipole is located at the cortex. As most of the brain function signals are located on the cortex, this means that tissue heterogeneity could cause about 20% RDM difference in most potential calculation cases. The general tendency of

MAG is similar to that of RDM, though much less dramatic. The MAG differences increase as the eccentricity increases, although it remains very close to 1 in all cases.

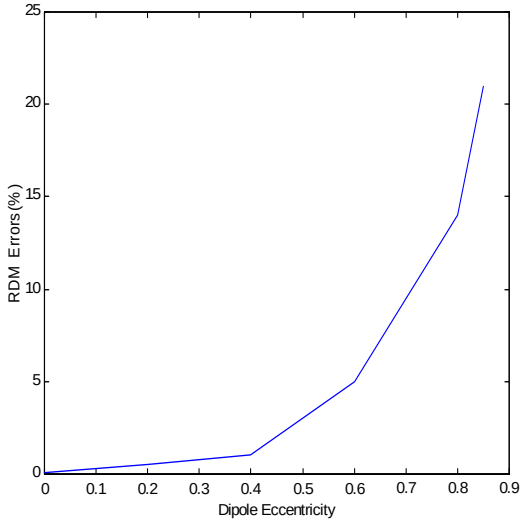

Figure 1. *RDM differences.*

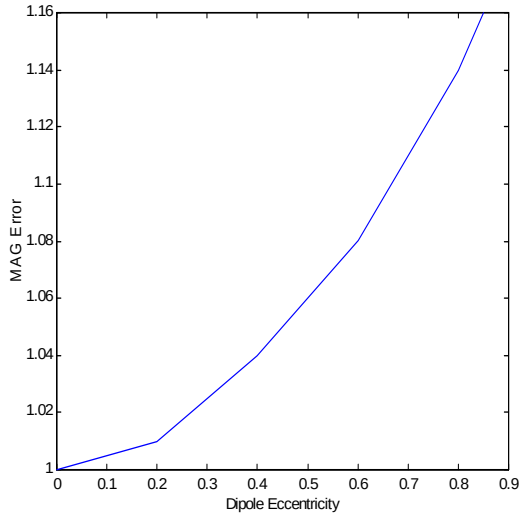

Figure 2. *MAG differences.*

Conclusions

This paper studies the heterogeneous head models that include the existing piecewise homogeneous models as a specific case where the pseudo-conductivity variance is equal to zero. Different heterogeneous head models were created and compared to the piecewise homogeneous head model through their computed potentials to experimentally explore the effect of these heterogeneity. To investigate the influence of each tissue on the computed potentials, the tissues modelled in both the heterogeneous head model and the piecewise homogeneous head model were changed from heterogeneous to homogeneous and vice versa. Based on these investigations, a guideline to include or exclude a wide variety of heterogeneities in a detailed model was given. This guideline allows researchers to make a trade-off between the model complexity and the number of computations and to make the modelling and computation problem manageable for a given problem. The influence of tissue heterogeneity on the anatomically accurate head model was also investigated. For this purpose, a head model with heterogeneous conductivity and another one with piecewise homogeneous conductivity were developed. The comparisons found that the differences introduced by tissue heterogeneity were significant, the conclusion being similar to that for the sphere models.

This study, in general, shows that the existing "realistic" head model is actually a geometrically accurate head model. The real realistic head model should include both the geometrical parameters and the conductive properties. The conductive heterogeneity within tissues has a significant influence on the potential distribution in a human head model.

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References

1. Plonsey, R. & Fleming, D., *Bioelectric Phenomena*, McGraw-Hill Book Company, New York, 1969.
2. Malmivuo, J. & R. Plonsey, R., *Bioelectromagnetism - Principles and Applications of Bioelectric and Biomagnetic Fields*, Oxford University Press, New York, 1995.
3. Awada, J. A. *et al*, "Computational aspects of finite element modelling in EEG source localisation", *IEEE Trans. Biomed. Eng.*, Vol. 44, No. 8, pp. 736 –752, 1997.
4. Cuffin, B. N., "EEG dipole source localisation: using inverse solution for determining source locations", *IEEE Engineering in Medicine and Biology*, pp. 118–122, Sep./Oct., 1998.
5. Gutierrez, D. & Muravchik C. H., "Estimating brain conductivities and dipole source signal with EEG arrays," *IEEE Trans. Biomed. Eng.*, vol. 51, pp. 2113–2122, Dec. 2004.
6. Duck, F. A., *Physical Properties of Tissue: A Comprehensive Reference Book*, Academic Press Limited, 24 - 28 Oval Road, London NW1 7DX, UK, 1990.
7. Oostendorp T. F., Delbeke J. & Stegeman, D. F., "The conductivity of the human skull: Results of in vivo and in vitro measurements," *IEEE Trans. Biomed. Eng.*, vol. 47, pp. 1487–1492, Nov. 2000.
8. Wen, P., *Human head modelling and computation for the EEG forward problem*, PhD thesis, Flinders University of South Australia, Adelaide, Australia, 2001.
9. Ferree, T.C., Eriksen K. J. & Tucker, D. M., "Regional head tissue conductivity estimation for improved EEG analysis", *IEEE Trans. Biomed. Eng.*, Vol. 47, No. 12, pp. 1584–1593, 2000.
10. Meijs, J. W. H. *et al*, "The influence of various heads on EEG's and MEG's", in *Functional Brain Imaging*, Pfurtscheller, G. & Silva, F. H. L., Eds., Toronto, Canada: Huber, pp. 31– 46, 1988.
11. Yan, Y., Nunez, P. L. & Hart, R. T., "Finite element model of the head: Scalp potentials due to dipole sources," *Med. Biol. Eng., Comp.*, Vol. 29, pp. 475– 481, 1991.

12. Lemieux, A. M. & Hand, J. W., "Calculation of electrical potentials on the surface of a realistic head model by finite differences", *Phys. Med. Biol*. Vol. 41, pp. 1079–1091, 1996.
13. Law, S. K., "Thickness and resistivity variations over the upper surface of the human skull", *Brain Topography*, Vol. 6, No. 2, pp. 99 –109, 1993.
14. Wen, P., He, F. B. & Sammut, K., "The heterogeneous conductivity property of cranial tissues and its representation in numerical head model", *Australasian physical & engineering sciences in medicine*, Vol.22, No. 3, pp. 92 – 98, 1999.
15. Goncalves S., de Munck, J. C., Verbunt J. P. A., Bijma F., Heethaar R., & da Silva F. L., "In vivo measurement of brain and skull resistivities using an EIT based method and realistic models for the head," *IEEE Trans. Biomed. Eng.*, Vol. 50, No. 6, pp. 754–767, June. 2003.
16. Ary, J. P., Klein, S. A. & Fender, D. H., "Location of sources of evoked scalp potentials: Corrections for skull and scalp thickness", *IEEE Trans. Biomed. Eng.*, Vol. 28, No. 6, pp. 447– 452, 1981.
17. Cuffin, B. N., "Effects of local variations in skull and scalp thickness on EEG's and MEG's," *IEEE Trans. Biomed. Eng.*, Vol. 40, pp. 42– 48, 1993.
18. Eshel, Y., Witman, S., Rosenfeld, M. & Abboud, S, "Correlation between skull thickness asymmetry and scalp potential estimated by a numerical model of the head", *IEEE Trans. Biomed. Eng*., Vol. 42, No. 3, pp. 242–249, 1995.
19. Haueisen, J, *et al*, "Influence of tissue resistivities on neuromagnetic fields and electric potentials studied with a finite element model of the head", *IEEE Trans. Biomed. Eng.*, Vol. 44, pp. 727–735, 1997.
20. Marin, G. *et al*, "Influence of skull anisotropy for the forward and inverse problem in EEG: simulation studies using FEM on realistic head models", *Human Brain Mapping*, Vol. 6, pp. 250–269, 1998.