The Influence of Pseudo Conductivities on the EEG Forward Computation for Human Head Modelling

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ABSTRACT

In brain imaging, the accuracy involved in calculating scalp potentials due to cerebral electric sources depends on the realism of the head model. Existing methods assume homogeneous conductivity throughout each component tissue. This assumption introduces inaccuracies in computing the potentials. This paper proposes a new approach based on the use of pseudo-conductivity values in place of the uniform-conductivity values assigned to tissues. Simulation results reveal that the conductivity values have a significant effect on the computed potentials, thereby invalidating the uniform-conductivity assumption.

1. INTRODUCTION

Electroencephalograms (EEGs) of scalp potentials provide a useful tool for non-invasively estimating the location of electric sources within the brain. These estimates are determined by comparing, for a given brain function, the recorded EEGs against the scalp potentials obtained from the computation of an electric field model of the head. These determinations are made by iteratively adjusting the source locations, strengths, and orientations in the head model, until the recorded and computed scalp potentials match each other. The accuracy of these estimated source locations depends not only on such factors as EEG measurement noise, EEG sensor location errors, computation errors, etc., but also, more importantly, on how closely the head model approximates the actual head [1-5]. There are two key factors which influence this approximation: (i) how closely the geometrical structure adopted by the head model represents the actual head structure, and (ii) how closely the electrical conductivity values assigned to each component cranial tissue for the computation of the electric field within the head model, match their real counterparts in the actual head. A brief summary of the effect due to (i) (i.e., geometric approximations) is given below. The effect due to (ii) (i.e., conductivity approximations) which has however not yet been discussed in the literature, forms the theme of this paper.

The earliest and simplest of the head models reported in the literature is a single homogeneous sphere of uniform electrical conductivity. Studies have shown that the use of this model produces large localisation errors because it does not account for the "smearing" effect on the scalp's EEGs due to the low conductivity of the skull [1-2]. Improvements in localisation accuracy have been achieved by using multilayer concentric spherical models in which the skull layer is characterised by a low conductivity value [3]. Due to their regular geometric structures, the computation of both the scalp potentials and the source locations for these concentric models can be implemented analytically. Significantly greater improvements in the computation results of the scalp potentials have been further achieved using more *'realistic'* geometric structures which closely approximate the shape of the head [4-10]. Due to their geometric irregularities, the scalp potentials for these realistic head models must be computed using numerical techniques, such as the Finite Element Method (FEM) and the Boundary Element Method (BEM). The localisation errors reported for the above techniques vary between 0.4cm and 3cm [8-10], depending on the location of the source within the brain, the realism of the geometric head model adopted, and the characteristics of the mesh structure used for the numerical computation.

For the numerical solutions of the concentric and the realistic head models, their geometric structures are meshed into many elements (e.g., 10^6) where each element is assigned the properties of a particular tissue type, including its characteristic conductivity. In the literature, the characteristic conductivity for each element of a tissue type is taken as the mean conductivity computed from several samples of that tissue type; thus all elements of the same tissue possess the same mean conductivity. In reality, however, due to the complex composite structure of the cranial tissues, the conductivities of a tissue type vary from one location to another within that same tissue type, with a distribution centred about a mean value. Since it is difficult to obtain the exact conductivity for each element corresponding to its location in the head, researchers have used the mean value as its characteristic conductivity. The use of the mean conductivity values, however, introduces a significant deviation from the actual head properties, thus resulting in inaccuracies in the calculations of the scalp potentials and the source localisations.

The paper suggests a new method referred to as pseudo conductivity head model, to investigate the influence of the conductivity on the calculation of the scalp potentials. Section

2 presents discussion of the conductivity properties of cranial tissues. Section 3 describes an investigative study to examine the influence of mean and pseudo conductivities on the computation results of a numerical head model. The final section discusses the simulation results and draws the concluding remarks.

2. ACTUAL AND PSEUDO CONDUCTIVITIES

Methods to improve the spatial resolution of EEG increasingly require the use of models which account for the actual conductivities and geometries of the cranial tissues [11-12]. Although dipole models using homogeneous three- and four-layer concentric and eccentric spheres still continue to be used for source estimation, significant errors in source localisation can result as a consequence of the disparities between these models' properties and the actual physical properties of the human head [11]. By using realistic head models, i.e., head models which closely approximate the actual conductive and geometric properties of the human head, it should be possible to reduce the error in source estimation algorithms. Consequently, recent efforts have been directed at developing models which properly reflect the true geometric and conductive properties of the human head. For example, Surface Laplacian (current density) and cortical mapping techniques improve spatial resolution in EEG by minimising the effect of volume conduction [13-16]. Existing finite element, finite difference, and boundary element methods improve spatial resolution through more accurate head geometry models [17-18].

As mentioned earlier, one aspect involved in accurately modelling the head, is to account for local variations in conductivity. Spherical head models use separate layers to represent scalp, skull, brain, and in some models, cerebral spinal fluid (CSF). In each layer, the thickness and conductivity along that layer are usually assumed to be uniform [10], [12], [16], [19-20]. Ideally, a uniform medium between source and sensor is desired because it can be analytically modelled and is computationally less intensive. In reality, however, neither the layer thickness nor its conductivity is uniform but can instead vary widely within the human head. Variations in the intervening medium between source and sensor can affect the flow of electric currents and alter the subsequent localisation estimates. Changing the conductivity and layer thickness in head models results in pronounced differences in source localisation estimates [10], [12], [20-21]. Thus, the use of realistic head models that more closely match local parameter variations across the real head, should allow more accurate and valid estimations of source locations by reducing the systematic errors introduced by previous simplifying assumptions concerning uniform parameters.

Ever since the discovery of bioelectric events, there has been an interest in the ability of living tissue to conduct current. Measurements of bio-conductivities appeared in the literature as early as 1902 for animal tissues and 1932 for human tissues. L. A. Geddes and L. E. Baker later compiled and published these data in 1967 [22]. Though numerous papers have been published from that time onwards, most of them took the above data as their basis. In the past two decades, further studies have been made by Chakkalakal et al, 1980 [23]; Kosterich et al, 1984 [24]; Woolley, 1986 [25] and Law, 1993 [26]. Law, in particular, described the procedure for measuring the conductivity of human skull tissue and concluded that the conductivity of human tissue varies with location even for the same type of tissue. At best the conductivities of bio-tissues can only be estimated but never properly measured using current techniques.

These observations have led many researchers to more comprehensively measure the specific conductivities of bio-tissues in order to better evaluate and provide some theoretical basis for determining the extent to which the volume conductor matrix can be considered homogeneous and in what manner inhomogeneity can affect recordings of electric potentials.

In the forward computation of EEG using FEM, the human head is modelled as a large number of elements; each representing a different area of the head. Different cranial tissues are represented as segments in realistic models and layers in concentric/eccentric models. Each segment or layer obviously has different conductivities. If however the simplifying assumption concerning uniform conductivities within tissues is removed, then each individual element will have its own unique conductivity. This is due to the complex composition of the tissue. For instance, elements in the brain may have different conductivities, since they may contain different proportions of blood vessels, white matter, grey matter, etc.. Experimentally measured values of conductivity for grey matter increase as a function of the measuring signal frequency (e.g., $0.33(\Omega m)^{-1}$@5Hz, $0.43(\Omega m)^{-1}$@5kHz, etc.). White matter has conductivity $1.76(\Omega m)^{-1}$@5Hz, and has been shown to be anisotropic with the ratio of conductivities varying between 5.7-9.4 [27]. The conductivity of the CSF surrounding the brain is generally accepted to be $1.0(\Omega m)^{-1}$. In the skull's case, the element conductivity may differ for elements composed purely of cancellous bone or compact bone, or some combination of the two. Its resistivity varies between 1360Ω-cm and 21400Ω-cm, with a mean of 7560Ω-cm and a standard deviation of 4230Ω-cm. All models reported in the literature use the value of $0.33(\Omega m)^{-1}$ for the scalp conductivity [22]. No allowance is made for the conductivity of the underlying muscle (0.0076-$0.52(\Omega m)^{-1}$), or subcutaneous fat (0.02-$0.07(\Omega m)^{-1}$) [28]. As a consequence of such variation in conductivity values within tissues, as well as the difficulty involved in measuring the exact corresponding conductivity for each element, the only feasible approach is to set the conductivity to some fixed representative value. Given that the conductivities of the elements for the same tissue are relatively close in comparison with those for different tissues, the conductivities of the elements in a tissue can therefore be assumed to follow a normal distribution:

$$f(x) = \frac{1}{\sigma\sqrt{2\pi}} e^{-\frac{1}{2}\frac{(x-\mu)^2}{\sigma^2}} \qquad (1)$$

where μ is the mean conductivity and σ the standard deviation of conductivity. The curve of $f(x)$ is symmetric with respect to $x = \mu$ because the exponent contains $(x-\mu)^2$. Changing μ corresponds to moving the curve to another position, which represents the uniform conductivity of the layer is modified. For small σ^2, we get a high peak and steep slopes, it means that, the conductivities of the elements within the tissue are tightly centred around the mean. And for $\sigma^2 = 0$, all elements in the tissue have the same conductivities; the mean conductivity - as assumed in the current literature. Conversely, with increasing σ^2, the conductivities of the elements are more widely distributed. From the assumption given in equation (1), a set of statistical parameters (namely, μ and σ^2) can be derived for a tissue type. A range of conductivity values - the *pseudo conductivities* - can then be generated to fit the Normal distribution which is specifically defined by μ and σ^2. These pseudo conductivities are allocated to the component elements belonging to that tissue. This pseudo-conductivity approach forms the basis of the technique proposed in this paper for introducing greater realism in head modelling.

3. NUMERICAL MODEL AND SIMULATION

The bioelectric fields produced in the human head can be mathematically described by Maxwell's equations. For electrostatic problems in dielectric volume conductors, an electric field, E, can be described in terms of the gradient of a scalar potential fields V,

$$E = -\nabla V \qquad (2)$$

In the frequency range of the EEG signals (0-100Hz), the volume conductor can be considered purely resistive, so that a quasistatic approximation is justified [14]. Then, the electric potential $V(x,y,z)$ that results from a current source $I(x,y,z)$ in a volume conductor can be described mathematically by Poisson's equation for electrical conduction.

$$\nabla(k\nabla V) + I = 0 \quad \text{in} \quad \Omega \qquad (3)$$

with boundary condition

$$\begin{aligned} V &= u(x,y,z) \quad \text{on} \quad S_1 \\ k\nabla V\vec{n} &= g(x,y,z) \quad \text{on} \quad S_2 \end{aligned} \qquad (4)$$

Where V is the electrostatic potential, k is the conductivity tensor, I is the current source, and S and Ω represent the surface and volume of the head.

FEM is used to solve the volume conductor problem described by the above equations numerically. The main strength of FEM is that it computes an estimate of the potential field around each element based on the material properties of that individual element. Therefore, it is possible to specify different conductivity tensors for each elements in different locations.

To evaluate the algorithm, a four-layer concentric spherical model of the head is considered. The four-layer model consists of a sphere with three concentric spherical shells, which correspond to the brain, the cerebrospinal fluid (CSF), the skull and the scalp. The outer radii of these layers are 7.9, 8.1, 8.5 and 8.8 cm, respectively. Each compartment has an average isotropic conductivity of 0.33, 1.0, 0.0042 and 0.33(Ωm)$^{-1}$, respectively.

A 3-D numerical model based on the above set of parameters is built with 20880 tetrahedral elements. The scalp, skull, and CSF shells are each allocated 5400 elements while the brain volume shell is allocated the remaining 4680 elements.

Since the concentric model used in this experiment is symmetric, the finite-element model can be reduced to a partial domain of the concentric model so long as the partial model preserves the original boundary conditions. Typically, a half-head model is used in applications where the computation of electric potentials is concerned.

The follow simulations are carried out using a semi-spherical model where the midline symmetric plane becomes the boundary surface of the FEM model. The simulation studies are conducted for both the uniform conductivity case and the pseudo conductivity case. Figure1 shows the conductivity values used in each element for the pseudo-conductivity case, while Figure 2 shows the corresponding conductivity values for the uniform-conductivity case. In the pseudo-conductivity case, the conductivities of each layer are normally distributed. The relevant parameters describing the distribution are listed in Table 1.

Table 1. Distribution parameters.

	μ (Ωm)$^{-1}$	σ^2
Brain Layer	0.33	0.1089
CSF Layer	1.00	1.00
Skull Layer	0.0042	1.764×10^{-5}
Scalp Layer	0.33	0.1089

For each conductivity case, two studies are carried to examine the influence of conductivity values on the electric potentials generated by single dipole sources and multi-dipole sources. These are described below.

Study 1: A single-radial dipole current source is placed at a distance of 7.5 cm from the centre of the sphere. Its location is (1.7533, 0.5284, 7.2747). Figures 3 and 4 show the potential distributions due to the single dipole.

Study 2: Multi-radial dipole current sources are respectively placed at (1.7533, 0.5284, 7.2747), (0.5284, 1.7533, 7.2747), and (-1.7533, 0.5284, 7.2747). Figures 5 and 6 shows the potential distribution due to these three dipoles.

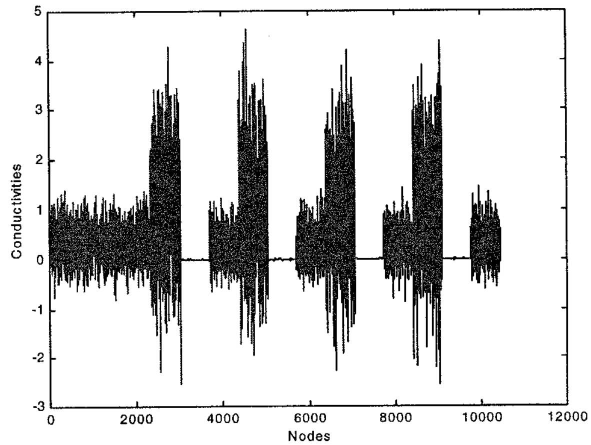

Figure 1. Pseudo-conductivity case.

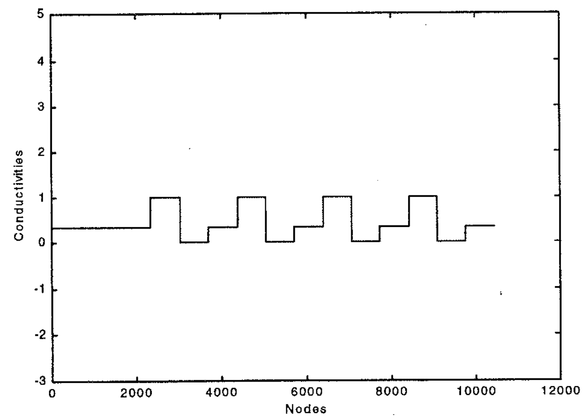

Figure 2. Uniform-conductivity case.

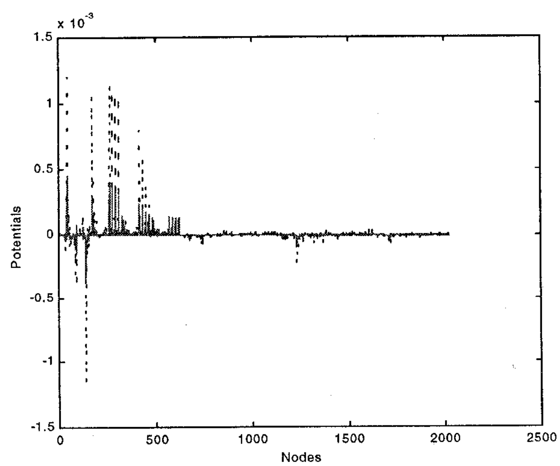

Figure 3. Single-dipole uniform-conductivity case.

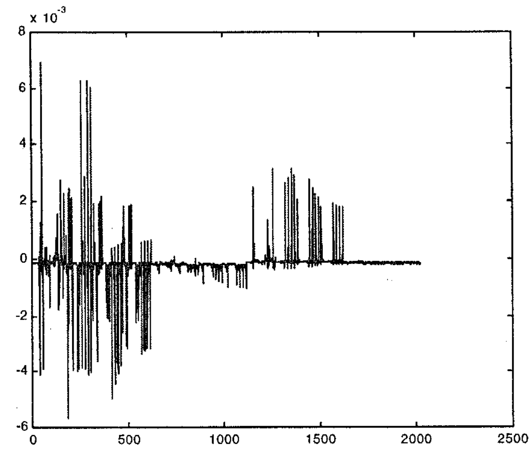

Figure 4. Single-dipole pseudo-conductivity case.

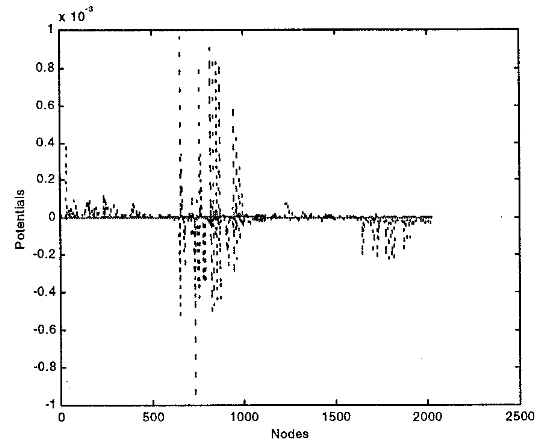

Figure 5. Multiple-dipole uniform-conductivity case.

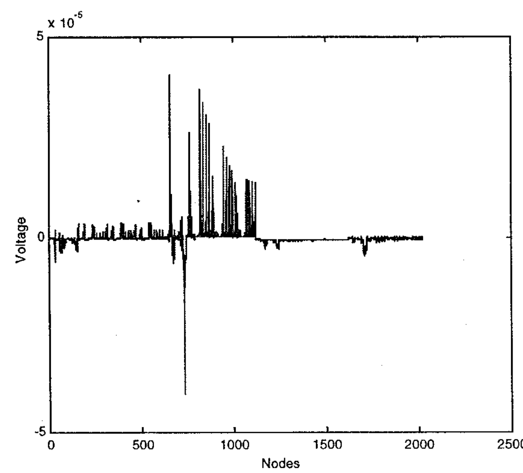

Figure 6. Multiple-dipole pseudo-conductivity case.

An analysis of these simulation results is conducted based on the comparison of the results of the uniform-conductivity case and the pseudo-conductivity case, using the following three criteria defined in (5), (6), and (7). Two of them, (5) and (6), are modified versions of the criteria used by Meijs *et al.* [29].

$$RDM = \sqrt{\int_\Omega \left(\frac{V_u}{\sqrt{\int_\Omega V_u^2}} - \frac{V_p}{\sqrt{\int_\Omega V_p^2}} \right)^2} \quad (5)$$

$$MAG = \sqrt{\frac{\int_\Omega V_p^2}{\int_\Omega V_u^2}} \quad (6)$$

$$V_{\max} = \max(\left|V_{u,i} - V_{p,i}\right|,\ i = 1,2,3.....) \quad (7)$$

The *RDM* qualifies errors in topography, whereas the *MAG* represents the magnification factor of the pseudo-conductivity solution V_p with respect to the uniform one V_u. Ideal values for RDM and MAG are thus respectively 0 and 1. The $V_{\max}$ gives the maximum difference between V_p and V_u. There are differences in these three measures (*RDM, MAG,* and $V_{\max}$) between those computed for the uniform-conductivity case and those computed for the pseudo-conductivity case, for both single-dipole and multi-dipole studies. Details of these differences in measures are listed in Table 2.

Table 2. Differences in measures.

	RDM	*MAG*	$V_{\max}$
Single Dipole	0.371	0.357	0.6552
Multiple Dipole	0.8259	0.0367	0.968

Both the simulation results shown in Figures 3-6 and the analysis presented in Table 2 show significant variations in potentials between those computed for the uniform-conductivity case and those computed for the pseudo-conductivity case. The values for *RDM* and *MAG* in both the single-dipole study and the multi-dipole study are far from their ideal values, namely 0 and 1 respectively. In all three criteria, the multi-dipole study is influenced more by the conductivity values than is the single-dipole study. The general tendency is that pseudo-conductivity approach produces larger magnitudes of potentials than does the uniform-conductivity approach.

4. DISCUSSION

The pseudo-conductivity approach presented in this paper represents a step forward in the construction of a mathematical model of the human head. It implements not only the inhomogeneity exhibited by different component cranial tissue types, but also the inhomogeneity exhibited by different elements within the same tissue type. The principle of this approach is based on the relaxation of the constraints imposed by the commonly-used uniform-conductivity approach, in order to allow the assignment of conductivities that are normally distributed as is characteristic of the actual human head.

The pseudo-conductivity approach makes full use of an important feature of FEM - its ability to account for individual material properties of each element. It also allows existing FEM algorithms to work more efficiently by interpolating element conductivities from the limited measured data set. The simulations presented in this paper are conducted on both the uniform- and pseudo-conductive models, for both single and multi dipole studies. The results show that, for both studies, the conductivity values have a significant impact on the computed potentials. This reveals that the uniform-conductivity assumption currently employed in existing head models is not accurate. Thus the potential distribution computed using the uniform-conductivity approach can not represent the real situation correctly. Further studies are being carried out to improve and verify this current approach in more detail.

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