A PSEUDO-CONDUCTIVITY INHOMOGENEOUS HEAD MODEL FOR COMPUTATION OF EEG

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Abstract: Human head models for the forward computation of EEG using FEM require a large set of elements to represent the head geometry accurately. Anatomically, the electrical property of each element is different, even though they may represent the same type of tissue (white matter, grey matter, etc.). Since it is impossible to obtain the electrical properties of the cranial tissues for every element in the head model, most algorithms which claim to deal with inhomogeneity can, in reality, only implement the computation for the homogeneous case. This paper presents a new numerical approach which can more precisely model the head by using a set of pseudo conductivity values for the computation of the inhomogeneous case. This set of values are extrapolated from the limited amount of real conductivity values which are available in the literature. Simulation studies, based on both this proposed approach and the homogeneous approach which utilises mean-valued conductivities, are performed. The studies reveal that the computation results for the potential distribution on the surface of the scalp, obtained using both approaches, are significantly different.

I. Introduction

It is commonly accepted that the mechanism underlying the generation of the electroencephalogram (EEG) can be physically described as a set of current sources embeded in a conductive medium [1]. When estimating the voltage distribution on the scalp resulting from dipole or distributed sources inside the head, a model is required which describes the dimensions of the conducting medium, the electrical conductivity of the medium, and the position and orientation of the sources. Since the head is approximately symmetrically spherical, sphere models have been developed in the literature to describe the head. These models of varying complexity include: the homogeneous sphere model, the 3- and 4-layer concentric models, the isotropic multi-layer concentric model, the anisotropic multi-layer concentric model [2], and the so-called realistic head model [3-6]. The simplicity of the calculations of the homogeneous model makes its use, together with the Ary [3] transformation of eccentricities, a fast and convenient method for modelling electrical sources in the brain, and it has been used in commercial packages such as BESA [4]. Recently, Zhang and Jewett [5] showed that using the homogeneous model to approximate the 3-layer concentric model leads to errors. Their results suggest that it can be dangerous to use oversimplified models. Previous methods for computing the 3- and 4-layer concentric models assume that the conductivity within a spherical shell is constant. In reality, however, due to the composite structure of cranial tissues, the conductivities at each point in the head are unique, even though they may be of the same tissue type. When the Finite Element Method (FEM) is used to compute the potential distribution on the scalp, the model must be meshed into a set of elements. Since the model may be composed of millions of elements, it is currently impossible to determine the exact conductivity for each element.

In this paper, a new numerical approach, which uses pseudo conductivity values rather than mean conductivity values to approximate the exact inhomogeneity of the cranial tissues, for the head model is presented. The paper shows how the pseudo conductivity values are derived from available statistical data. The correlation between the conductivities and the potential distribution on the scalp is analysed by using a 4-layer concentric FEM model with mean and pseudo conductivities, respectively. The aims of the paper are to demonstrate how an inhomogeneous head model can be generated from limited medical data, and to show that the potential distribution on the scalp obtained using the proposed approach varies significantly from that obtained using the mean-valued approach.

II. Method

The current trend in head modelling is to develop anatomically accurate models of the head. The most detailed model reported in the literature to date is, in our best knowledge, that of Cuffin [6]. This model uses magnetic resonance images to obtain a 3-D field of data points, and then employs harmonic expansion to interpolate between the data points, thereby creating a reduced set of mesh verities. The verities are then connected with linear triangular elements to create the desired surface. In Cuffin's method, one of the two limiting factor for quantitative analysis of the head model – the anatomical accuracy – can be reasonably approximated, whereas the second limiting factor – the local variation in conductivity – is not addressed. Both the concentric and realistic head models mentioned previously use separate layers to represent the scalp, skull, brain, and in some models, the cerebral spinal fluid (CSF). The conductivity across each layer is usually assumed to be a

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fixed mean value. In reality, however, the conductivity varies within a layer, because the composition of the associated tissue can vary widely with location. Furthermore, variations in layer thickness of the intervening medium between the source and the sensor can affect the current flow. Changing the conductivities or the layer thickness in the head model can generate pronounced variations in the localisation of electric sources in the brain [6-7]. Much effort has been made in the attempt to obtain the electrical properties of living tissues.

Sets of conductivity data for bio-tissues have been reported in the literature as early as 1902 [8-9]. Initially, they were limited to animal tissues. In 1967, Geddes and Baker [9] published the first compendium of human tissue conductivities based on work formerly reported in 1932. Though a number of related papers have been published since then, most of them used Geddes and Baker's data as their basis. In the past two decades, further studies have been made by Chakkalakal et al [10], Kosterich et al [11], Woolley [12], and Law [13]. In particular, Law's paper described the details for determining the conductivity of the human skull, and concluded that the conductivities of human tissues differ from one location to another, even for the same tissue. In most cases, the tissue conductivity can, at best, only be estimated.

In the forward computation of EEG using FEM, the human head is modelled by a large number of elements; each representing a different area of the head with its own unique conductivity. Not only do the elements representing different tissues have unique conductivities, but so do the elements representing the same type of tissue. The latter is due to the complex composition of the tissue. 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.. 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 [14]. 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 [9]. No allowance has been made for the conductivity of the underlying muscle (0.0076-$0.52(\Omega m)^{-1}$) [15], or subcutaneous fat (0.02-$0.07(\Omega m)^{-1}$) [15]. With such widely varying values of conductivity, the only feasible approach is to set the conductivity to a fixed mean value. It is impossible (or at least not easy) to measure and set an exact conductivity for each element. 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 Gaussian 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 is the standard deviation. The curve of $f(x)$ is symmetric with respect to $x=\mu$ because the exponent contains $(x-\mu)^2$. Changing μ corresponds to translating the curve to another position. σ^2 is the variance. For small σ^2, the conductivities of the elements within a tissue are tightly centred around the mean, and for $\sigma^2=0$, all conductivities are the same – 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) can be derived for a tissue type from the limited data available for that tissue in the literature. A range of conductivity values – the *pseudo conductivities* – can then be generated to fit the Gaussian distribution which is specifically defined by μ and σ. These pseudo conductivities are allocated to the component elements belonging to that tissue.

III. Results

To evaluate the proposed approach, a 4-layer concentric head model is considered. This model consists of four concentric spherical shells, which correspond to the brain, the CSF, the skull, and the scalp, respectively. The corresponding radii of the surfaces of these shells are 7.9, 8.1, 8.5, and 8.8 cm, respectively. Each shell has a mean isotropic conductivity of 0.33, 1.0, 0.0042, and $0.33(\Omega m)^{-1}$, respectively.

A 3-D numerical model based on the above parameters is built using 20880 tetrahedral elements. In the scalp, skull, and CSF volumes, each shell has 5400 elements, respectively. The remaining 4680 elements are used to fill the brain volume.

The follow simulations are carried out based on the assumption that the spherical model is symmetric, thus only half of the model needs to be computed. The simulation results both for the mean-valued conductivity case and the pseudo conductivity case are presented. To verify the correlation between conductivity and potentials, the variance σ^2 of the distribution is varied between 0 and μ^2. When $\sigma^2=0$, the model regresses to the case of mean-valued conductivity. As the value of σ^2 increases, the conductivity range for each shell also increases. Each element in the model can then be assigned a pseudo conductivity value within the range as defined by σ^2 and μ. Fig. 1 presents the pseudo conductivity distribution ($\sigma^2=\mu^2$) and Fig. 2

presents the mean-valued conductivity distribution ($\sigma^2 = 0$). The numbers of elements in each shell are listed in Table 1.

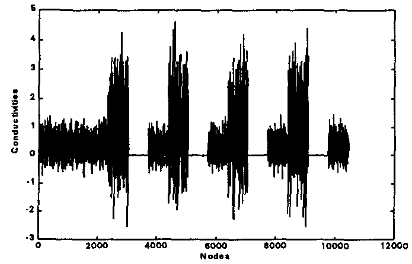

Fig. 1. Pseudo conductivity distribution ($\sigma^2 = \mu^2$).

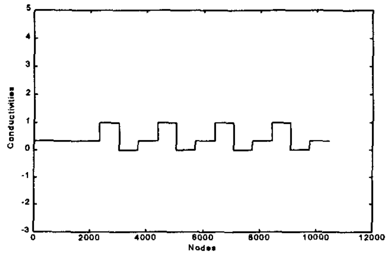

Fig. 2. Mean-valued conductivity distribution ($\sigma^2 = 0$).

Table 1: Number of Elements in Each Shell.

	Brain	CSF	Skull	Scalp
No. of Elements	2340	2700	2700	2700

The aim of the following simulation study is to determine, for given conditions, the influence of pseudo conductivity on the computed potential distribution results. A comparison between the results of the mean-valued conductivity case and the pseudo conductivity case is made using three criteria ((4)-(6)), two of which ((4)-(5)) were introduced by Meijs et al [16]:

$$RDM = \sqrt{\int_{\Omega}\left(\frac{V_u}{\sqrt{\int_{\Omega} V_u^2}} - \frac{V_p}{\sqrt{\int_{\Omega} V_p^2}}\right)^2} \qquad (4)$$

$$MAG = \sqrt{\frac{\int_{\Omega} V_p^2}{\int_{\Omega} V_u^2}} \qquad (5)$$

$$V_{\max} = \max(|V_{u,i} - V_{p,i}|,\ i = 1,2,3.....) \qquad (6)$$

The RDM quantifies the errors in topography, whereas the MAG represents the magnification factor of the pseudo conductivity solution (V_p) with respect to the mean-valued conductivity solution (V_u). Ideal values for RDM and MAG are 0 and 1, respectively. The $V_{\max}$ represents the maximum difference between V_p and V_u.

FEM is used for computing the potential distribution for the 4-layer concentric model for both conductivity cases. A quasistatic formulation and linear media were assumed. A single radial dipole source is placed 7.5cm from the centre of the spheres. The co-ordinates of the dipole are: (1.7533, 0.5284, 7.2747). Fig. 3 and Fig. 4 show the potential distributions due to the single dipole for both the mean-valued conductivity case ($\sigma^2 = 0$) and the pseudo conductivity case ($\sigma^2 = \mu^2$), respectively. The RDM, MAG and $V_{\max}$ measures using (4)-(6) are given in Table 2.

Table 2: The RDM, MAG and $V_{\max}$ measures for $\sigma^2 = 0$ to μ^2.

σ^2	RDM	MAG	$V_{\max}$
0	1	0	0
$(0.1\mu)^2$	1.0187	0.0372	2.81E-5
$(0.2\mu)^2$	1.5877	0.0705	2.57E-4
$(0.3\mu)^2$	1.0751	0.1255	1.11E-4
$(0.4\mu)^2$	1.416	0.8649	0.002
$(0.5\mu)^2$	1.4712	0.5951	8.214E-4
$(0.7\mu)^2$	1.5298	0.7167	0.0021
μ^2	8.3471	1.3851	0.0067

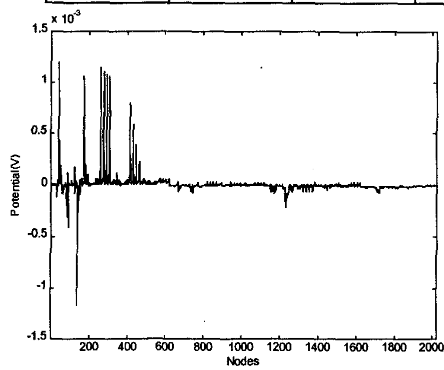

Fig. 3. Potential distribution for $\sigma^2 = 0$.

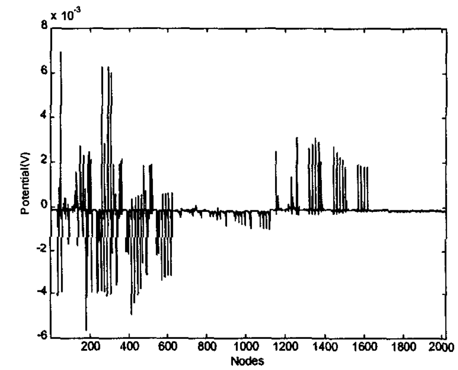

Fig. 4. Potential distribution for $\sigma^2 = \mu^2$.

IV. Discussions

The simulation results suggest that local variations in conductivity within elements have a non-negligible effect on the computation of potential distributions in the head. When σ^2 is increased from 0 to μ^2, the potential amplitude increases, and the change becomes significant when σ^2 is greater than $(0.5\mu)^2$. It is found that the change in potentials for the extreme case of $\sigma^2=\mu^2$, relative to the potentials obtained for $\sigma^2=0$, is profound.

The RDM and MAG measures show the effects of changes in conductivity more clearly. When $\sigma^2<(0.3\mu)^2$, RDM and MAG are almost equivalent to their ideal values (1 and 0, respectively), whereas when $\sigma^2>(0.5\mu)^2$, both RDM and MAG increase significantly away from their ideal values. This effect can also be observed in the V_{max} measure. As σ^2 increases from 0 to $(0.3\mu)^2$, V_{max} changes from 0 to 1.11×10^{-4} (v) which is negligible compared to the maximum potential (1.2×10^{-3} (v)) for $\sigma^2=0$. However, for $\sigma^2>(0.3\mu)^2$, V_{max} increases rapidly as σ^2 increases, and for $\sigma^2=\mu^2$ the value of V_{max} is 6.7×10^{-3}(v) which is significantly larger than the maximum potential for $\sigma^2=0$. This implies that the effect of inhomogeneity of the head model on the potential distribution calculation can not be ignored, and that the assumption of mean-valued conductivity adopted by existing head models is not accurate.

V. Conclusion

This paper has proposed a novel approach for the calculation of the potential distribution in the human head based on a 4-layer concentric model. The novelty of the proposed approach is in the use of pseudo conductivity values in place of mean conductivity values for the generation of the inhomogeneous head model. The pseudo conductivity values are derived from Gaussian distributions of the statistical conductivity data available in the literature. The use of pseudo conductivity more closely mimics the inhomogeneity of the human head, thus allowing a more realistic method for modelling the head..

Simulations conducted using both the mean-valued conductivity approach and the pseudo conductivity approach indicate that the effect of variance in conductivity for a bio-tissue within the head is indeed significant. This is particularly true, when the variance in conductivity is close to the mean square value.

The study shown in this paper reveals that the assumption currently used in existing head models – that conductivities for elements representing the same tissue type can be set to a mean value for that tissue – is not accurate. Thus the potential distribution computed using mean-valued conductivity is not reliable. Further studies are being carried out by the authors to confirm the validity of the pseudo conductivity approach.

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