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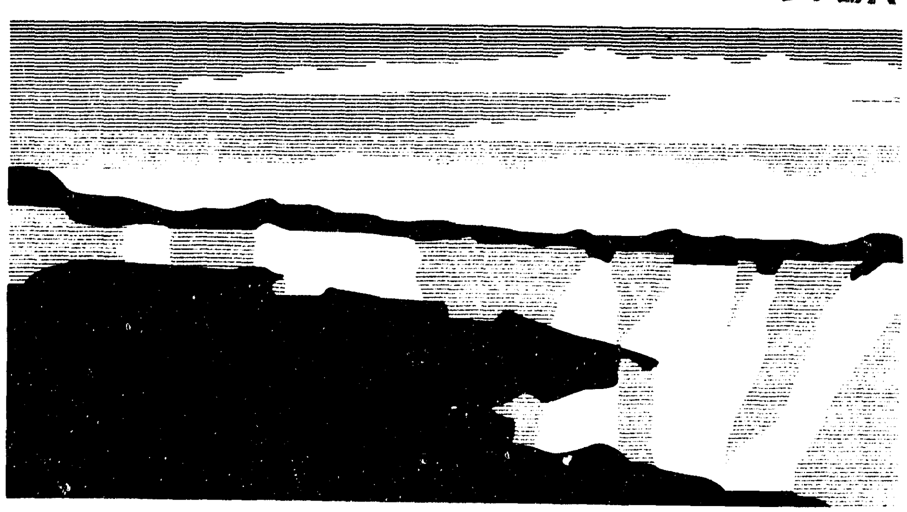

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Effect of Conductor Geometry on Source Localization: Implications for Epilepsy Studies

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Abstract

We shall discuss the effects of conductor geometry on source localization for applications in epilepsy studies. The most popular conductor model for clinical MEG studies is a homogeneous sphere; its popularity is due primarily to its simplicity. The geometry of a sphere is simple to describe, and the magnetic field due to a dipole is straightforward to calculate. However, several studies [1, 2, 3, 4] have indicated that a sphere is a poor model for the head when the sources are deep, as is the case for epileptic foci in the mesial temporal lobe. We believe that replacing the spherical model with a more realistic one in the inverse fitting procedure will improve the accuracy of localizing epileptic sources.

In order to include a realistic head model in the inverse problem, we must first solve the forward problem for the realistic conductor geometry. Analytic solutions do not exist for calculating the magnetic field due to current sources in a realistic head model, so it is necessary to compute the electric potential numerically. We create a conductor geometry model from MR images, and then solve the forward problem via a boundary integral equation for the electric potential due to a specified primary source. Once the electric potential is known, the magnetic field can be calculated directly. The most time-intensive part of the problem is generating the conductor model; fortunately, this needs to be done only once for each patient. It takes little time to change the primary current and calculate a new magnetic field for use in the inverse fitting procedure.

We present the results of a series of computer simulations in which we investigate the localization accuracy due to replacing the spherical model with the realistic head model in the inverse fitting procedure. The data to be fit consist of a computer generated magnetic field due to a known current dipole in a realistic head model, with added noise. We compare the localization errors when this field is fit using a spherical model to the fit using a realistic head model. Using a spherical model is comparable to what is usually done when localizing epileptic sources in humans, where the conductor model used in the inverse fitting procedure does not correspond to the actual head.

1 Procedure

1.1 Magnetic Field Calculation

In a spherical model, analytic solutions for the electric potential and magnetic field exist. In a realistic head model, on the other hand, analytic solutions do not exist and it is necessary to compute the electric potential numerically. Only after the electric potential has been calculated, can the magnetic field be found.

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The electroencephalography (EEG) forward problem consists of computing the electric potential that is produced by any assumed primary current, $\mathbf{J}^p$. In the brain the primary current flows within neurons, and is the quantity of interest in neuroscience. This current flows across the cell membrane and throughout the electrically conducting extracellular medium, where it is called the "return" current. The total current $\mathbf{J}$ is the sum of these two parts, and in the quasistatic approximation can be written

$$\mathbf{J} = \mathbf{J}^p - \sigma \nabla V \tag{1}$$

where σ is the electric conductivity and V is the electric potential.

For a head model in which the potential is a (different) constant in each of several compartments, the condition that $\mathbf{J}$ be continuous, $\nabla \cdot \mathbf{J} = 0$, can be transformed into an integral equation for V [5]

$$\frac{\sigma_i^- + \sigma_i^+}{2} V(\mathbf{r}) = V^p - \frac{1}{4\pi} \sum_j (\sigma_j^- - \sigma_j^+) \int_{S_j} dS' \mathbf{n}(\mathbf{r}') \cdot \nabla' \frac{1}{|\mathbf{r} - \mathbf{r}'|} V(\mathbf{r}'), \tag{2}$$

where V^p is the potential due to $\mathbf{J}^p$ in a conductor of infinite extent and unit conductivity

$$V^p = \frac{1}{4\pi} \int d^3 r' \mathbf{J}^p(\mathbf{r}') \cdot \nabla' \frac{1}{|\mathbf{r} - \mathbf{r}'|}. \tag{3}$$

The sum in Equation 2 runs over all the surfaces at which σ changes value; $\mathbf{r}$ is a point on surface S_i and $\mathbf{n}(\mathbf{r}')$ is a unit outward normal vector at the point $\mathbf{r}'$.

The main task in obtaining a numerical solution of Equation 2 is to find a good approximation to the integral on the right hand side in terms of the values of the unknown function V at some discrete set of points. We use the approach for solving the boundary integral equation proposed by de Munck [6], in which the electric potential varies linearly across each plane triangle of the mesh. We have found [7, 8] that this approach gives accurate and reliable results for spherical conductor geometries.

The Boundary Element Method suffers from poor accuracy on the outer two surfaces when the skull conductivity is much smaller than the scalp and brain conductivities. To overcome the loss of numerical significance, we employ the isolated-problem approach developed by Hämäläinen and Sarvas [4].

After solving Equation 2 for V a further numerical integration yields the magnetic field at any point outside the head [9]

$$\mathbf{B}(\mathbf{r}) = \mathbf{B}^p - \frac{\mu_0}{4\pi} \sum_j (\sigma_j^- - \sigma_j^+) \int_{S_j} dS' \mathbf{n}(\mathbf{r}') \times \frac{\mathbf{r} - \mathbf{r}'}{|\mathbf{r} - \mathbf{r}'|^3} V(\mathbf{r}'), \tag{4}$$

where $\mathbf{B}^p$ is the magnetic field due to the primary current.

1.2 Mesh Generation

Practical applications of the procedures we propose require efficient procedures for segmentation of volumetric MRI or CT data. Obtaining a mesh representation of a head surface (Figure 1) from an MRI data volume requires setting a voxel (3-D pixel) intensity threshold corresponding to scalp, and "shrinkwrapping" an icosahedral mesh to voxels having values above this threshold. The inner and outer skull surfaces are more difficult to obtain, requiring segmentation of the appropriate objects in the MRI data volume before a mesh shrinkwrapping is done.

2 Methods

We present a series of computer simulations in which we generate magnetic fields for a particular forward model, add noise, then fit the noisy data using a variety of conductor models. We consider only single time slices, and only one dipole at a time.

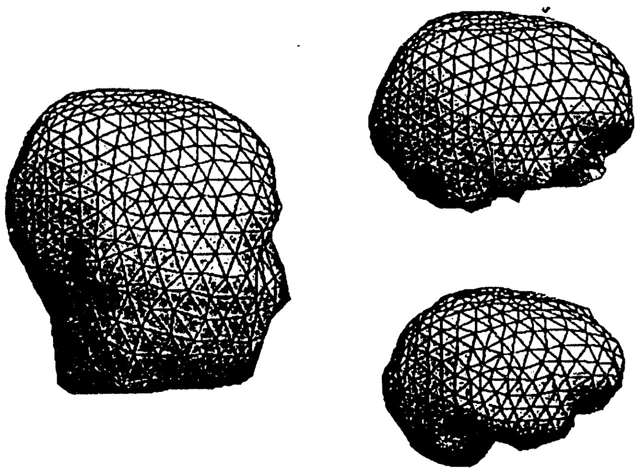

Figure 1: Meshes representing the scalp (left hand side), skull (upper right), and brain (lower right) surfaces. Each mesh is made up of 1280 plane triangles.

The data to be fit is generated by calculating the magnetic field due to a single dipole in a 3-layer (brain, skull, and scalp) model of a human head. The conductivity ratios are $\sigma_{brain} = \sigma_{scalp} = 100\ \sigma_{skull}$. Gaussian random noise is added to the magnetic field to make the Signal to Noise Ratio (SNR) 10:1, where SNR is defined as the ratio of the peak signal to the standard deviation of the noise.

The sensor geometry consists of 127 sensors on a sphere of radius 12 cm which is centered at (0, 0, 2 cm) (Figure 2). This sensor arrangement covers most of the area of interest, although it does not capture both magnetic field peaks for the deepest dipole. Each sensor is a first-order gradiometer with baseline of 5.1 cm, and coil radius 1 cm. Figure 2 shows the sensors as disks surrounding the inner skull surface. The dark shading on the inner skull surface represents the best fitting sphere surface.

A total of seven different dipole locations (Table 1) are considered, one at a time. All dipole orientations are tangential with respect to the best fitting sphere. The magnitude of the deepest dipole is 240 nA-m, all other dipoles have magnitude of 200 nA-m.

Table 1: True dipole locations in cm.

DIPOLE	X	Y	Z	$R = \sqrt{(x^2 + y^2 + z^2)}$
a	1.0	2.4	2.4	3.53
b	1.0	3.6	2.4	4.44
c	-0.7	2.4	3.7	4.46
d	-1.7	2.4	4.9	5.71
e	1.0	5.6	2.4	6.17
f	-3.1	2.4	6.8	7.84
g	-4.5	2.4	8.6	9.99

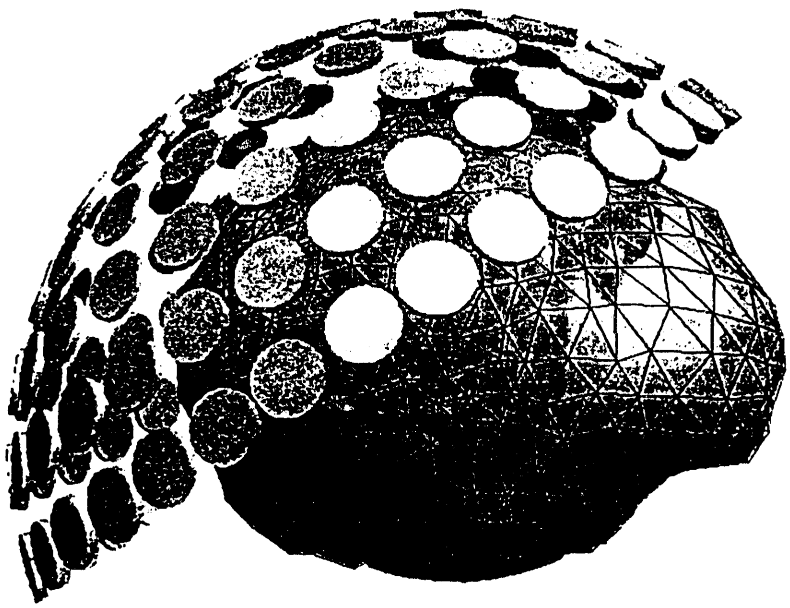

Figure 2: Sensor geometry, mesh on inner skull, and best fitting sphere.

The noisy data is fit using 3 different inverse models: 1) a 3 layer model of the head (the same geometry that was used to generate the original data), 2) a single surface representing the inner skull surface, and 3) a sphere that best fits, in a least-squares sense, the inner skull surface. The best fitting sphere has a radius of 6.8 cm, and is centered at (0.25, 0, 5.26 cm). Each surface is approximated by 1280 plane triangles, and the conductivities of the brain and best-fitting sphere are the same as the brain conductivity in the 3-layer head model.

3 Results

We use a non-linear least squares fitting routine to find the location, strength and orientation of each dipole in turn. We input the true dipole location as a starting value for the inverse procedure. The results of the fits are shown in Table 2. When the inverse model was the 3-layer head, the dipole fits were better than using either the inner skull or the best fitting sphere. The poor fits for the deeper dipoles (dipoles a through d) indicate that the conductivity geometry used to fit data is important. The error in location can be over 2 cm for deep dipoles, if the best fitting sphere is used for the inverse model. In contrast, the location error is less than 0.5 cm (for all dipoles studied) when the 3-layer head model is used. Using only the inner skull for the inverse model results in better dipole fits than with the best fitting sphere, but worse fits than with the 3-layer head. The inverse model is not as critical for the shallower dipoles (dipoles e through g). For the shallower dipoles, the average location error was 0.23 cm for the 3-layer head model, 0.4 cm for the inner skull model, and 0.5 cm for the best fitting sphere.

The disadvantage of using a realistically shaped conductor model, either a 3-layer head model or just the inner skull, is an increase in cpu time for the dipole fitting routine. For example, it took about 20 minutes of cpu time on a HP 735 to fit the deepest dipole using the 3-layer head, 10 minutes using the

Table 2: Error in dipole fit locations in cm for the three inverse models used in this study.

	INVERSE MODEL		
DIPOLE	3-Layer Head	Inner Skull	Best Fitting Sphere
a	0.22	0.59	2.59
b	0.21	0.97	1.44
c	0.21	0.36	2.19
d	0.07	0.25	1.57
e	0.34	0.71	0.93
f	0.07	0.14	0.55
g	0.25	0.30	0.13

inner skull, and about 30 seconds using the analytic equations for a sphere. The cpu time to create the realistic conductor model, which needs to be done only once for each patient, is around 5 to 10 minutes, depending on the complexity of the model. We consider the increase in time to be worth the increase in localization accuracy.

4 Acknowledgments

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