Technical report

An evaluation of dipole reconstruction accuracy with spherical and realistic head models in MEG

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Abstract

MEG forward problem has been solved for about 2000 dipoles placed on the brain surface using a very fine 3-layer realistic model of the head and the boundary element method (BEM). For each dipole, spherical models, one-layer realistic BEM models and coarser 3-layer realistic BEM models, were used to reconstruct the dipole. It was found that the localization bias induced by using a spherical model of the head increased from 2.5 mm in the upper part of the head to 12 mm in the lower part, on average. It was also found that, for the same computing time, a 3-layer model of the head gave on average 2 mm better localization errors than a one-layer model of the head. Orientation errors of less than 20° could only be retrieved with a 3-layer realistic model. Localization and orientation errors highly depended on the dipole position in the brain.

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1. Introduction

The spherical head model is widely used for estimating sources of the electrical brain activity from magnetoencephalographic signals (MEG). The general belief that the spherical model is more accurate for MEG than for electroencephalographic signals (EEG) stems from the fact that MEG does not depend on conductivity ratios between layers in a spherical geometry. The bias induced on dipole reconstruction by the spherical approximation has been systematically tested for magnetic data by Fuchs et al. (1998), but only using a 1-layer realistic model of the head.

More sophisticated spherical models consisting of eccentric spheres have been proposed in MEG by Meijs and Peters (1987), and further studied by Cuffin (1991) in terms of varying skull and scalp thicknesses. The difference in localization accuracy between the concentric sphere model and these models was found to be negligible (0.3 mm error). However, only one tangential dipole placed on an axis of symmetry of the model geometry was considered.

More realistic head models have been introduced by Hämäläinen (1987), and Meijs et al. (1987). Magnetic resonance images (MRI) segmentation is used to obtain meshes for real-shaped skull, scalp and brain, and the Boundary Element Method (BEM) can then be applied to solve the forward problem. Since magnetic field is less sensitive to conductivity distribution, it is often considered that a model with only the brain layer is sufficient. In Zanow (1997), the localization error obtained on data generated by a one-layer realistic model with a spherical inverse model was ranging from 1 mm for superficial dipoles near the vertex to 25 mm for deeper frontal dipoles. In Tomita et al. (1996), a similar simulation gave localization errors ranging from 0.67 mm for a shallow occipital dipole to 21 mm for the deepest frontal dipole. The orientation error for almost tangential dipoles was ranging from 1 to 63°. However, 12 of the 15 dipoles had orientation errors less than 10°. In Fuchs et al. (1998), a systematic simulation study with about 3600 dipoles and a one-layer realistic model for generating the data showed that the localization error using a spherical model ranged from 0.4 to 45.1 mm for tangential dipoles in a 122 channels helmet system. Smaller localization errors where obtained for tangential dipoles located near the pick-up coils.

Beside those simulation studies, the spherical model accuracy was also tested experimentally in MEG, using realistic phantoms of the head, or implanted dipoles in epileptic patients undergoing presurgical intracerebral recordings.

In Barth et al. (1986), a human cadaver head in which the brain had been replaced with salt jelly was used. Five
dipoles and 81 sequentially recorded positions for the single MEG sensor were used. Localization errors obtained with a spherical model ranged from 1.3 mm for shallow sources to 9.5 mm for the deepest sources. In Menninghaus et al. (1994), a non-conducting plastic skull filled with a saline solution was used, corresponding to a one-layer physical model of the head. Localization errors ranging from 1.3 mm for shallow sources to 9.5 mm for the deepest sources. In Menninghaus et al. (1994), a human skull was impregnated and filled with saline gelatin, and coated with latex to obtain a 3-layer physical model of the head. Magnetic fields successively generated by 32 dipoles (left central sulcus, left calcarine fissure, and frontal area) were recorded with a Neuromag-122 planar gradiometer. Using a spherical model, and the R-MUSIC localization method, they found errors ranging from 1 to 17 mm. The BEM, based on realistically deformed spherical meshes, lead to a 0.5-mm improvement of the localization error only.

In Cohen et al. (1990); Cohen and Cufn (1991); Balish et al. (1991), MEG localization errors using implanted dipoles were found to be ranging from 3 to 20 mm, with a mean varying from 8 to 10 mm, strongly depending on the dipole depth, and provided that the signal-to-noise ratio

$$\text{SNR} = \frac{\langle B(t)^2 \rangle_{\text{dipole on}}}{\langle B(t)^2 \rangle_{\text{dipole off}}}$$

was greater than 10. However, SNR was acceptable for only a few dipoles, possibly because most dipoles had a radial orientation with respect to the inner skull surface. In Rose et al. (1991) 3 implanted dipoles in the infero-temporal lobe were localized within 1–4 mm.

The localization errors found in these simulation and experimental studies using the spherical head model for dipole reconstruction are summarized in Table 1. However, most of these MEG studies considered very few dipole locations.

Yvert et al. (1997) reported a systematic evaluation of the dipole localization accuracy obtained from electrical signals with a spherical model. The work presented here aims at extending this study to the magnetic case.

For a given 143 MEG sensor configuration, magnetic fields were computed with a very fine realistic 3-layer BEM model for 2003 dipoles distributed on the cortical surface. Then, the localization bias due to the use of a spherical head model, a one-layer realistic BEM model, or a coarse 3-layer realistic BEM model were studied. Dipole orientation errors were also systematically evaluated in the case of realistic geometries. This aspect was less frequently reported in the literature (Tomita et al., 1996; Leahy et al., 1998).

This MEG simulation study focused on the evaluation of the localization bias due to different level of geometrical approximations of the fine reference model. The influence of other sources of errors such as background recording noise or coil location uncertainties are beyond the scope of this paper.

### Table 1

<table>
<thead>
<tr>
<th>Authors</th>
<th>MEG data</th>
<th>Number of dipoles</th>
<th>Dipole depth (mm)</th>
<th>Dipole orientation</th>
<th>Localization error (mm)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Cufn (1991)</td>
<td>Simulated data, eccentric spheres model</td>
<td>1</td>
<td>31</td>
<td>Tangential</td>
<td>0.3</td>
</tr>
<tr>
<td>Zanow (1997)</td>
<td>Simulated data, one-layer realistic model</td>
<td>30</td>
<td>7–53</td>
<td>Tangential</td>
<td>1–25</td>
</tr>
<tr>
<td>Tomita et al. (1996)</td>
<td>Simulated data, one-layer realistic model</td>
<td>15</td>
<td>≈ 15–55</td>
<td>Tangential</td>
<td>0.7–21.4</td>
</tr>
<tr>
<td>Fuchs et al. (1998)</td>
<td>Simulated data, one-layer realistic model</td>
<td>≈ 3600</td>
<td>≈ 0–80</td>
<td>1/3 Tangential, 1/3 intermediate, 1/3 radial</td>
<td>0.4–45.1, 0.2–70.9, 13.1–87.9</td>
</tr>
<tr>
<td>Barth et al. (1986)</td>
<td>Human cadaver</td>
<td>5</td>
<td>≈ 20–75</td>
<td>Variable</td>
<td>1.3–9.5</td>
</tr>
<tr>
<td>Menninghaus et al. (1994)</td>
<td>Phantom, one-layer</td>
<td>4</td>
<td>10–30</td>
<td>Tangential</td>
<td>3–9</td>
</tr>
<tr>
<td>Leahy et al. (1998)</td>
<td>Phantom, 3-layer</td>
<td>32</td>
<td>10–41</td>
<td>Variable</td>
<td>1–17 (SNR &lt; 4)</td>
</tr>
<tr>
<td>Cohen et al. (1990)</td>
<td>Implanted dipoles</td>
<td>12</td>
<td>≈ 30–45</td>
<td>10 Radial, 2 tangential</td>
<td>4–40 (4–13 for SNR &gt; 10)</td>
</tr>
<tr>
<td>Rose et al. (1991)</td>
<td>Implanted dipoles</td>
<td>3</td>
<td>≈ 15–45</td>
<td>Tangential</td>
<td>1–4</td>
</tr>
</tbody>
</table>

### 2. Methods

#### 2.1. Mesh generation

Meshes for the BEM were derived from a stack of MRI slices following a 3-step method. First, from each MRI slice, scalp, skull and brain contours were extracted. Second, contours were meshed together following the method described in Yvert et al. (1995) leading to very refined meshes (about 20 000 nodes). Third, mesh density was...
reduced according to the simplification technique proposed by Algorri and Schmitt (1996) in order to obtain a tolerable number of triangles for practical use (Table 2).

2.2. Forward model

The simulated MEG data were computed using the BEM with linear interpolation of the potential on the surface elements (de Munck, 1992; Ferguson et al., 1994) and the isolated problem approach (Meijs et al., 1989; Hämäläinen and Sarvas, 1989). The model used for this computation (called REF model in the following) had about 4500 nodes and a high triangle density (mean: 3 triangles/cm²) (Table 2) and is shown in Fig. 1. The normalized conductivity values were 1, 1/80, 1 for the scalp, skull and brain compartments respectively, as suggested in Rush and Driscoll (1968); Homma et al. (1995).

A linear interpolation method being used for the BEM, a key feature of the meshes considered in this study is the number of nodes which determines directly the size of the matrix equations to be solved.

2.3. MEG sensor configuration

MEG data were computed on a whole-head sensor configuration from the CTF company (143 first order gradiometers, with a baseline coil 5 cm away from the pickup coils). This arrangement was realistically positioned around the REF model (Fig. 1). The distance between scalp and sensors was as much constant as possible (minimum distance: 19 mm).

2.4. Choice of the dipole sites

The 2003 dipole sites were identical as those used in Yvert et al. (1997), enabling further comparison between EEG and MEG results. They were randomly chosen from the intersection of a cortical surface obtained by MRI segmentation and 10 horizontal planes separated by 10 mm (Fig. 2). The distance between two neighboring dipoles within a plane was about 5 mm. Each dipole orientation was normal to the cortical surface at the dipole location. The dipole depth under the cortical surface was ranging from 0.9 to 46.9 mm (mean ± 16.4 ± 9.8 mm). As shown in Fig. 2, deeper dipoles lead to magnetic fields having a lower root mean square value (RMS).

To characterize the dipole orientation, a ‘tangentiality’ angle (between 0 and 90°) was defined as the angle between the dipole moment and its radial component in the global sphere model (GLSPH, described in Section 2.5.1). In the presentation of the results, dipoles having a tangentiality angle between 0 and 5° will be referred to as radial dipoles, whereas those with a tangentiality angle between 85 and 90° will be referred to as tangential dipoles. Of the 2003 dipoles, 63% have a ‘tangentiality’ angle between 60 and 90°. Fig. 2 gives an overview of the tangentiality over all 2003 dipoles.

2.5. Inverse models

2.5.1. The spherical models

Four spherical models were considered for solving the inverse problem:

1. A global sphere model (GLSPH, Fig. 1) which best fitted the surface of the head-shape covered by all coils.
2. A brain-fitted global sphere model (BRAINGLSPH), similar to GLSPH but fitting the brain surface instead of the head shape.
3. A local sphere model (LOCSPH, Fig. 1) which best fitted a surface of the head-shape near the coils where the signal had its minima and maxima. This surface was determined as follows:
   (a) The distance d between the pickup coils recording the minimum and maximum signals was computed.
   (b) Then, a subset of coils whose distance from the middle of those two coils was less than 2 * d was selected. If more than half of the total number of coils were selected, only the global sphere was considered, and the dipole was discarded from the comparison between spherical inverse models.
   (c) The local sphere best fitted the surface of the head-shape whose distance from the middle of those two coils was less than 2 * d.
4. A brain-fitted local sphere model (BRAINLOCSPH), similar to LOCSPH but fitting the brain surface instead of the head shape.

This procedure for selecting a subset of coils and for

<table>
<thead>
<tr>
<th>Model name</th>
<th>Model type</th>
<th>Number of layers</th>
<th>Number of nodes per layer</th>
<th>Triangle density (tri/cm²) scalp-skull-brain</th>
</tr>
</thead>
<tbody>
<tr>
<td>Forward problem</td>
<td>REF</td>
<td>Realistic</td>
<td>3</td>
<td>1500</td>
</tr>
<tr>
<td>Inverse problem</td>
<td>GLSPH</td>
<td>Global best-fitting sphere</td>
<td>–</td>
<td>–</td>
</tr>
<tr>
<td></td>
<td>LOCSPH</td>
<td>Local best-fitting sphere</td>
<td>–</td>
<td>–</td>
</tr>
<tr>
<td></td>
<td>1L900</td>
<td>Realistic</td>
<td>1</td>
<td>900</td>
</tr>
<tr>
<td></td>
<td>3L300</td>
<td>Realistic</td>
<td>3</td>
<td>300</td>
</tr>
<tr>
<td></td>
<td>3L900</td>
<td>Realistic</td>
<td>3</td>
<td>900</td>
</tr>
</tbody>
</table>
defining a local sphere has been chosen because it could usually be applied in practice, when a head-shape discretization is available: it does not require any a priori assumption on the possible location of the sources. Spheres fitting the head-shape are usually chosen only when no MRI is available. In the other case, brain fitted spheres are usually preferred.

Apart from those four spherical models, the sensitivity of spherical models to sphere center position was tested using six sphere models shifted from BRAINLOCSPH by 1 cm in the 6 axial directions.

2.5.2. The one-layer realistic model

From a topographical analysis of forward problems solved with one-layer and 3-layer models for a few source configurations, Hämaäinen and Sarvas (1989) have suggested that since the skull has a very high resistivity compared to scalp and brain tissues, i.e. major currents are flowing in the brain, only one layer is needed to model the head. Here, we extended this study by systematically testing this assumption for about 2000 dipoles. We used a one-layer model (Fig. 1, Table 2) having about 900 nodes and a density of 2.5 triangles/cm².

2.5.3. The 3-layer realistic models

We also considered two realistic 3-layer models: one coarse (3L300), and one fine (3L900) having respectively 300 and 900 nodes per layer (see Fig. 1, Table 2). The most time-consuming steps in the inverse computations being matrix inversion and matrix product, 3L300 and 1L900 had similar computational time.

2.6. The inverse procedure

The inverse problem was solved iteratively by varying the dipole position and orientation parameters $\mathbf{p}$ so as to minimize the following criterion:

$$
\varepsilon = \sum_{i=1}^{N} \left( B_{i}^{\text{REF}} - B_{i}^{\text{INVmodel}(p)} \right)^{2}
$$

where N is the number of channels, and $B_{i}^{\text{REF}}$ is the magnetic field computed with the REF model at the $i^{th}$ channel, and $B_{i}^{\text{INVmodel}(p)}$ is the magnetic field produced at the $i^{th}$ channel by a dipole with parameters $\mathbf{p}$ for a given inverse model.

A mixed linear/non-linear iterative algorithm (Marquardt) was used for the minimization (Scherg, 1990; Verkindt et al., 1995), which required initial conditions for the dipole parameters. In this study, the exact position of the

![Fig. 1. Simulation procedure. Models used for the forward and inverse procedure are presented with superimposed pickup coils positions. REF is a 3-layer realistic model with a triangle density of 3 tri/cm², and is used for creating the simulated MEG data. GLSPH and LOCSPH are spherical models used for inverse computations. 1L900 is a one-layer realistic model with 2.6 tri/cm², 3L300 and 3L900 are 3-layer models with 0.7 tri/cm² and 1.8 tri/cm² respectively.](image-url)
Fig. 2. Top: A 3D view of the segmented cortical surface has been superimposed on a mid-sagital MRI slice. The projection of the dipoles positions on the segmented cortical surface are represented by red dots. Dipoles are placed on 10 horizontal slices spaced by 10 mm. Middle: Root Mean Square values (RMS) of the magnetic field created by each dipole. The dipole population is divided into 4 equally numbered parts depending on their associated RMS values. For each dipole position, a color dot is superimposed on MRI slices according to the RMS value for that dipole: Red, 0 fT < RMS < 0.5 fT; yellow, 0.5 fT < RMS < 0.9 fT; green, 0.9 fT < RMS < 1.3 fT; blue, 1.3 fT < RMS < 14 fT. Bottom: Tangentiality values. For each dipole position, a color dot is superimposed on MRI slices, the color being chosen according to the tangentiality of that dipole: red, 0–30° (almost radial); yellow, 30–60°; green, 60–90° (almost tangential).
original dipole was taken as the initial conditions. For 186 shallow dipoles, the 3L300 mesh was so rough that these dipoles fell outside the meshed brain surface, and were thus discarded from the statistical analysis. Beside this, for 50 dipoles, the relative residual error (ε divided by the mean square value of the signal) at the end of the inverse procedure was over 30%, indicating that no reasonable solution was found. These cases, which mostly happened with spherical models, were also discarded.

2.7. Error criteria

The localization error D is defined as the distance in millimeters between the reconstructed dipole and the original dipole, the orientation error θ as the angle between the reconstructed dipole moment and the original dipole moment, and the moment amplitude error ΔM as the difference between the reconstructed dipole moment amplitude and the original dipole moment amplitude. ΔM is expressed as a percentage of the original dipole moment amplitude. Negative and positive values indicate underestimation and overestimation, respectively of the dipole strength.

3. Results

3.1. Bias due to the spherical approximation

For 824 dipoles (depth usually inferior to 2 cm), the local sphere (LOCSPH) and the global sphere (GLSPH) were different. For all these dipoles the brain fitted local sphere (BRAINLOCSPH) and the brain fitted global sphere (BRAINGLSPH) were also different. The distance between the centers of the scalp-fitting spheres ranged from 2.5 to 58 mm (mean 14.7 ± 6.9 mm). For comparison of GLSPH, BRAINGLSPH, LOCSPH and BRAINLOCSPH, only these 824 dipoles were considered.

3.1.1. Localization errors

We found that for those 824 dipoles, GLSPH, LOCSPH, BRAINGLSPH and BRAINLOCSPH lead to a mean localization error of 3.5, 2.4, 3.1 and 2.5 mm, respectively, in the upper part of the head (slices 1–5 in Figs. 3A and 4). The 1.1 mm difference between GLSPH and LOCSPH was significant (paired t test: \( P < 0.0001 \)), and so was the 0.6 mm difference between BRAINGLSPH and BRAINLOCSPH. The 0.1 mm difference between LOCSPH and BRAINLOCSPH was not significant (\( P = 0.29 \)).

In the lower part of the head (slices 6–10), GLSPH, LOCSPH, BRAINGLSPH, and BRAINLOCSPH lead to increased mean localization errors of 5.3, 8.7, 4.6 and 4 mm, respectively. This vertical slice effect was significant (\( P < 0.0001 \)). A more detailed slice analysis in this region showed that:

- In the posterior part of the brain, all spherical models excepts BRAINLOCSPH gave a mean localization error between 3.7 and 3.9 mm: there was no statistical difference between these models (\( P > 0.6 \)). BRAINLOCSPH gave a mean localization error of 2.9 mm which was statistically different from other models (\( P < 0.0001 \)).
- ii) In the anterior part of the brain, GLSPH, LOCSPH, BRAINGLSPH, and BRAINLOCSPH lead to a mean localization error of 6.5, 13.4, 5.2 and 5 mm respectively. All those differences were significant (\( P < 0.0001 \)) except between BRAINLOCSPH and BRAINGLSPH (\( P = 0.62 \)).

When varying the BRAINLOCSPH sphere center in the six axial directions, localization errors varied on average of 0.8 mm. No particular direction gave systematically better results.

3.1.2. Orientation errors

Orientation errors were very similar for all spherical models (Figs. 3A and 5), and no obvious slice effect could be seen. It should be noted that each slice included a combination of dipoles with various orientations. As expected, radial dipole leads to orientation errors of 90°, whereas this error fell to 10° for tangential dipoles (Fig. 3B).

3.1.3. Moment amplitude errors

As for orientation errors, no obvious slice effect could be seen on moment amplitude errors (Fig. 3A). As expected, radial dipoles were localized with a moment amplitude error of −100%, whereas this error fell between −15 and 7% for tangential dipoles (Fig. 3B). All spherical models behaved similarly, except BRAINLOCSPH which lead to a significantly improved moment amplitude reconstruction (\( P < 0.007 \)).

3.2. Errors obtained with realistic models

All statistics included in this part were computed with the 1768 dipoles which fell inside the meshed brain layer for all meshes considered and gave a residual fitting error below 30%. Results were compared to those obtained with GLSPH and BRAINLOCSPH.

3.2.1. Localization errors

Localization errors obtained with the different realistic models (1L900, 3L300 and 3L900) are given in Figs. 4, 6A and 7A.

- For 1L900 (Fig. 7A), the mean localization errors increased from 2.4 mm in the upper part of the head (slices 1–5) to 4.9 mm in the lower part of the head (slices 6–10). They varied from 8.3 to 2.5 mm for radial and tangential dipoles respectively (Fig. 6A).
- For 3L300 (Fig. 7A) the mean localization errors increased from 1.3 mm in the upper and middle part of the head (slices 1–8), to 3.5 mm in the lower part of the
head (slices 9–10). They varied between 4 and 0.97 mm for radial and tangential dipoles respectively (Fig. 6A).

- For 3L900 (Fig. 7A) the mean localization errors increased from 0.4 mm in the upper and central part of the head (slices 1–8), to 1.4 mm in the lower part of the head (slices 9–10). They varied from 2.6 to 0.4 mm for radial and tangential dipoles respectively (Fig. 6A).

All these differences between lower and upper parts of the head were statistically significant (t test: \( P < 0.0001 \)).

Dipoles were localized better than 3 mm in 59, 90, and 98% of the cases with 1L900, 3L300, and 3L900, respectively (Fig. 4).

### 3.2.2. Orientation errors

Orientation errors obtained with the different realistic models (1L900, 3L300 and 3L900) are given in Figs. 5, 6B and 7B.

- For 1L900 the mean orientation errors remained between 13 and 31° when the dipole tangentiality varied.
- For 3L300 the mean orientation errors remained between 3 and 21° when the dipole tangentiality varied.
- For 3L900 the mean orientation errors remained between 2 and 8° when the dipole tangentiality varied.

Dipole orientations were retrieved with an accuracy better than 10° for 27, 60 and 87% of the dipoles with 1L900, 3L300, and 3L900, respectively (Fig. 5).

### 3.2.3. Moment amplitude error

Moment amplitude errors obtained with the different realistic models (1L900, 3L300 and 3L900) are given in Figs. 6C and 7C.

- For 1L900 the mean moment amplitude errors varied between −50 and 0% when the dipole tangentiality varied from 0 to 60°, and between 0 and 24% when the dipole tangentiality varied from 70° to 90°.
- For 3L300 the mean moment amplitude varied between −27 and 0% when the dipole tangentiality varied from 0 to 50°, and between 0 and 6% when the dipole tangentiality varied from 60 to 90°.
- For 3L900, the mean moment amplitude varied between −11 and 0% when the dipole tangentiality varied from 0 to 60°, and between 0 and 2% when the dipole tangentiality varied from 70 to 90°.

Dipole moment amplitudes were retrieved with an accuracy between −10 and 10% for 32, 67, and 87% of the dipoles with 1L900, 3L300, and 3L900, respectively.
Fig. 4. Localization error for BRAINLOCSHP and realistic models. (Note that 1L900 and 3L300 have the same computing cost). For each dipole position, a color dot is superimposed on MRI slices, the color being chosen according to the localization error obtained for that dipole: red: 0–3 mm; yellow: 3–6 mm; green: 6–12 mm; blue: >12 mm.
Fig. 5. Orientation error for spherical and realistic models. For each dipole position, a color dot is superimposed on MRI slices, the color being chosen according to the orientation error obtained for that dipole: a red dot indicates an error between 0 and 10°, a yellow dot, between 10 and 20°, a green dot, between 20 and 40°, and a blue dot, over 40°.
4. Discussion

The goal of the present MEG simulation study was to estimate the localization bias induced by different approximations of a given realistic reference model (REF). In particular, this work aimed at determining the brain regions where a spherical model provides similar results to those obtained with realistic models, and at determining the influence of realistic model characteristics on source position, orientation, and strength reconstruction. It is very likely that implanted dipole studies (phantoms or epileptic patients) provide the best realistic data for evaluating source localization methods. However, for practical limitations, only a few tens of dipole positions are generally considered (Barth et al., 1986; Cohen et al., 1990; Balish et al., 1991; Rose et al., 1991; Menninghaus et al., 1994; Leahy et al., 1998). By contrast, in simulation studies, like the present one, one has to choose a particular mathematical model of the head to generate the data set. However, they offer a mean for systematically evaluating localization errors all over the brain.

4.1. Localization bias due to spherical model

We have found that the localization bias induced by using a spherical model of the head ranged from about 2.5 mm in the uppermost part of the head up to about 8 mm in the temporal lobe. This ‘vertical position effect’ has also been observed in EEG using simulated data (Yvert et al., 1997), which results are shown here for comparison in Fig. 7. Similar important errors in the lower part of the head have been recently reported using spherical models for localization of equivalent sources underlying interictal spike discharges in epileptic patients (Merlet et al., 1998, Fig. 4). This effect is much less pronounced with realistic models both in EEG and MEG. The spherical model induces a smaller localization bias in the most spherical-like regions of the head which are also well covered by whole head helmet systems. The range of localization errors is similar to what Zanow (1997); Fuchs et al. (1998) found when using a one-layer model of the head for the simulated forward computation, and agrees with the range of errors found with artificial dipoles in epileptic patients (Balish et al., 1991; Cohen et al., 1990; Rose et al., 1991) or phantoms (Barth et al., 1986; Menninghaus et al., 1994). At first sight, our results seem in contradiction with those of Leahy et al. (1998), who found that a very refined 3-shell BEM model gave localization errors only 0.5 mm better than a local sphere, on average. However, their dipoles were distributed only along trajectories mimicking the central sulcus and upper occipital cortex, corresponding to brain areas where spherical and realistic models lead to similar errors in our simulation. Nevertheless, one of their dipoles was located in the lowest part of the head (dipole #12), for which the BEM model gave an error of about 4 mm, while the sphere model leads to about 17 mm error (Fig. 11 in Leahy et al., 1998).
For the other dipoles, which were located upper in the head, realistic and spherical models lead to comparable errors. Hence, our simulations are consistent with their results.

We also compared local and global sphere models. We found that for scalp-fitting spheres, the local sphere was slightly better (1 mm) than the global sphere in the upper part of the head. However, the situation was reversed in the lower part of the brain (14 mm for LOCSPH and 7 mm for GLSPH). By contrast, brain-fitted local spheres gave a significantly improved accuracy of 0.6 mm compared with brain-fitted global sphere. The local sphere has been systematically considered in the literature by authors who were using MEG devices with sensors partially covering the head-shape (BTI 37 channel system for instance). This approach was mainly used to explore the somatosensory and auditory cortices, in which areas we have found slightly
smaller errors with the scalp-fitting local sphere. In more recent studies (Mäkelä et al., 1994; Yvert et al., 1998) based on whole-head MEG systems, authors used a local sphere fitting a part of head shape covered by a subset of coils selected from a priori knowledge on the dipole location. Such choice might lead to important errors (over 8 mm) for sources in the infero-temporal or frontal regions, except if a brain-fitted local sphere is used. However, this sphere can only be chosen when MRIs are available, which is not often the case.

4.2. Influence of the number of layers with realistic models.

We have shown that the number of layers in BEM models may have an important influence on MEG single dipole reconstruction. We have found that the 3-shell model 3L300 leads to errors 2 mm smaller on average than with the one-shell model 1L900 for the same computing time. Moreover, as discussed below, the 3-layer models allow a much better reconstruction of the dipole orientation and moment amplitude. This result may be mainly explained by a better account for secondary current sources with a 3-layer model. It should be recalled however that our data are not real but simulated with a 3-layer BEM model of the head. Hence, whether the very often proposed one-layer realistic model is an oversimplification of the head geometry should be further investigated in more realistic situations such as implanted dipoles in epileptic patients.

4.3. Orientation and moment amplitude errors

We found that the dipole orientation and moment amplitude can be well estimated with realistic models of the head. It is commonly stated that MEG is not sensitive to the radial components of the dipoles. Our study showed that in realistic cases, the magnetic field distribution partially depends on this radial component, and that this information is sufficient (at least, in noiseless situations like here) to allow an estimation of the dipole orientation and moment amplitude. Fig. 6B shows that while the ability of the spherical model to reconstruct the dipole orientation and moment amplitude depends strongly on the dipole tangentiality (very poor for radial source), this is surprisingly not the case with realistic models. Indeed, for radial dipoles, the dipole orientation is reconstructed within less than 20° with both 3-layer realistic models, and 40° with the one-layer realistic model. The moment amplitude is retrieved with less than 15% error for radial dipoles and 3L900. For tangential dipoles the spherical models reconstruct the orientation of the dipoles within 10°. In this particular case it is even better than realistic models, most likely because the dipole is constrained to remain tangential, while the dipole can vary freely in orientation with realistic models. Fig. 7 shows that 3-layer BEM models retrieve the dipole orientation within comparable errors between EEG and MEG. Although deep and radial dipoles are often thought to be magnetically silent, in our data and others (Haueisen et al., 1995; Menninghaus and Lütkenhöner, 1994), such sources still produce a RMS signal value being 20% of that produced by a tangential source. Hence, the ability of realistic model to retrieve radial and deep sources should be further investigated in noisy simulations, which was beyond the scope of this study. Nevertheless, we have here considered the error results obtained for only the 75% of the dipoles leading to highest RMS values (Fig 7). It appears that this subset of dipoles lead to a similar behavior as the entire population. Thus, localization and orientation errors found in this study do not stem from low RMS dipoles only, but most likely from the dipole position and tangentiality in the head, and from geometrical approximation.

References

Meijis J, Bosh F, Peters M, Lopes da Silva. F. On the magnetic field distribution generated by a dipolar current source situated in a realistically