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The Dependence of the Inverse Solution Accuracy in Magnetocardiography on the Boundary Element Discretization

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***Abstract*--Modeling in magnetocardiography (MCG) is increasingly based on the boundary element method (BEM). We quantify the influence of the boundary element discretization on the cardiomagnetic forward and inverse problem for different dipole depths and regions of the heart. Simulations using single current dipoles and a high resolution BE model (edge length < 10 mm) are used to assess models of various complexity (with and without blood masses) and discretization. It is found, that the maximum localization error of about 5 mm occurs if the test dipole is very close to one of the boundaries (lungs). Edge lengths of 20, 15, and 8 mm for the torso, lungs, and ventricles, respectively, are sufficient to reach a localization accuracy of 2 mm.**

***Index Terms*—Biomagnetics, Boundary Element Methods, Cardiography, Inverse Problems.**

I. Introduction

The solution of the electromagnetic inverse problem plays an important role in the reconstruction of cardiac electrical activity from magnetocardiographic (MCG) measurements. Naturally, the accuracy of the inverse solution is greatly dependent upon the adequacy of the employed forward model. The boundary element method (BEM) is widely used to approximate the electrical conductivity profile of the human torso (e.g. [1,2]). This modeling techniques employs two important simplifications in order to facilitate the computational handling: (1) the conductivity is assumed to be isotropic and piecewise homogeneous, (2) the boundaries between compartments with different conductivity are discretized by triangles (boundary elements). With respect to the latter, it is an important question how the discretization (size of the triangles) influences the accuracy of the forward and the inverse solution. This question is tackled by this study. A representative selection of possible dipoles in the human myocard were assumed as sources for a set of simulated magnetocardiographic recordings (simulated magnetic field maps). These reference data sets were computed using a very fine boundary element model. Then, the inverse problem was solved from these data sets, using coarser or less complex (fewer boundaries) models. Additionally, the influence on the forward computation was assessed by comparing the simulated magnetic field maps computed with the coarser and less complex models with the reference data sets.

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II. Methods

The magnetic field due to a current dipole in a piecewise homogeneous and isotropic volume conductor was computed using the boundary element method with linear potential approximation and isolated potential approach [3]. The models were constructed out of a T1 weighted MRI data set of the torso of a healthy volunteer. Thirteen different models were constructed: model 0 through 6 consisting of five compartments (torso boundary, lung boundaries and ventricular blood mass boundaries), and model 7 through 12 containing only torso and lungs. All boundaries were thinned, and triangulated with side lengths of the triangles according to Table I. The maximal number of points for the reference model (BEM model 0) was limited by the computer available (standard desktop computer, 320 Mbyte RAM).

TABLE I
BEM Models

BEM No.	triangle side length in mm			points/triangles
	torso	lungs	ventricles	
0	10	6	3	7962 / 15608
1	12	8	4	5164 / 10198
2	14	10	6	3579 / 7084
3	16	12	8	2637 / 5220
4	20	15	8	1782 / 3526
5	25	20	8	1186 / 2340
6	25	20	10	1110 / 2188
7	10	6	-	6770 / 13326
8	12	8	-	4485 / 8870
9	14	10	-	3229 / 6374
10	16	12	-	2416 / 4788
11	20	15	-	1561 / 3094
12	25	20	-	965 / 1908

A homogeneous conductivity of 0.2, 0.04 and 0.6 S/m (torso, lungs, ventricular blood masses) was assumed. All source localizations were carried out with the help of the software CURRY V3.0 (NeuroScan, Sterling, VA, USA). Fig. 1 shows the simulation setup. A sensor configuration with 64 channels was employed for the simulation of the magnetic fields. We calculated the fields for each single dipole (d 1 to d 13) with each BEM model (model 0 through model 12). Then, we performed source localizations using the forward computed field map from BEM model 0 and all other volume conductor models (model 1 through model 12). Additionally, we employed the forward computed field map from BEM model 7 for source localizations with the BEM models not including the ventricles (BEM 8 to BEM 12). The localization error was computed as Euclidian distance between the dipole used in the forward simulation and the inverse solution. The strength difference between both dipoles was determined. We calculated the angle α between both dipoles according to $\cos\alpha = \vec{n}_{orig} \cdot \vec{n}_{calc}$, where $\vec{n}_{orig}$ is the original dipole orientation and $\vec{n}_{calc}$ is the inversely estimated dipole orientation.

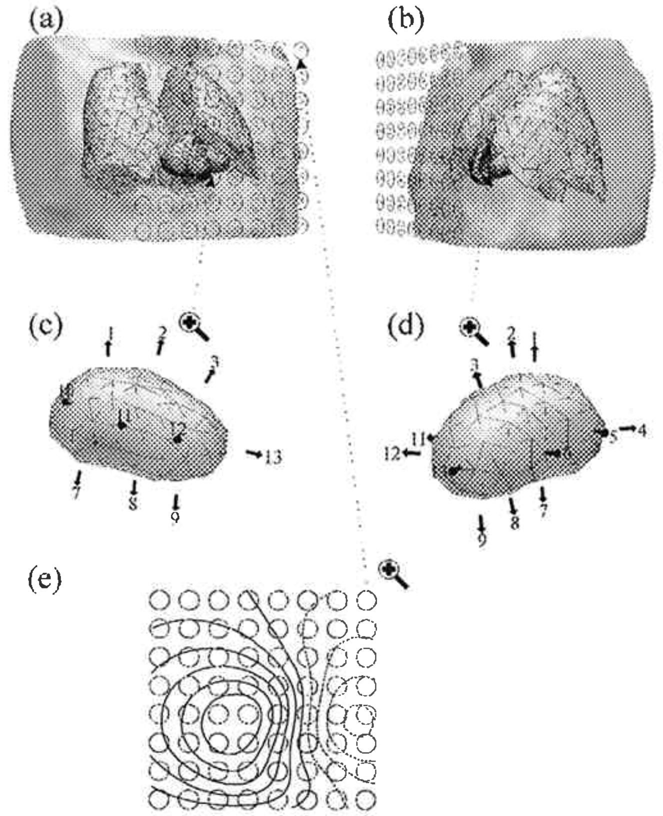

Fig. 1. Torso, lungs, and ventricular blood masses from a right anterior (a) and left anterior view. Enlarged view right (c) and left (d) anterior view on the left ventricle and locations for dipoles d1 to d13. Simulated magnetic field profile (e).

III. Results

Fig. 2 and 3 depict the localization error of the inverse computations for all BEM models. The error caused by the different BEM discretizations was not more than 3 mm except for dipole 5 in the coarsest BEM models (models 5, 6, and 12). These source localization errors are comparable to the errors found in our previous study on BEM discretization errors in human magnetoencephalography [4]. When omitting the ventricles in the modeling process, a larger localization error of up to 7 mm (Fig. 2, BEM model 7) was found. The outliers in the very coarse models can be explained by the small distance these dipoles have to the closest boundary. In these cases the minimal distance of 0.5 triangle side length (distance between dipole and closest triangle) was not reached.

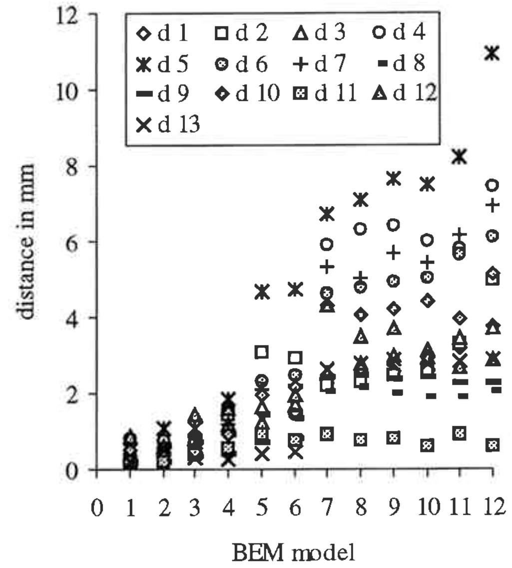

Fig. 2. Distance between original source location and inversely computed source location for all BEM models. BEM model 0 represents the reference model.

Fig. 4 shows the relative strength of the reconstructed dipoles. A relative strength of 1.0 corresponds to the actual strength of the original dipoles. The deviation from the original dipole strength is almost 20 percent, except for two outliers (40 % change, dipole 5, BEM model 5 and 6, not shown in the figure).

Fig. 5 indicates the angle α between the original dipole and the reconstructed dipoles. The error in the dipole orientation produced by the coarser discretizations is up to 10 degrees, while the model simplification was found to produce orientation errors of almost 40 degrees (Fig. 5, dipole 10).

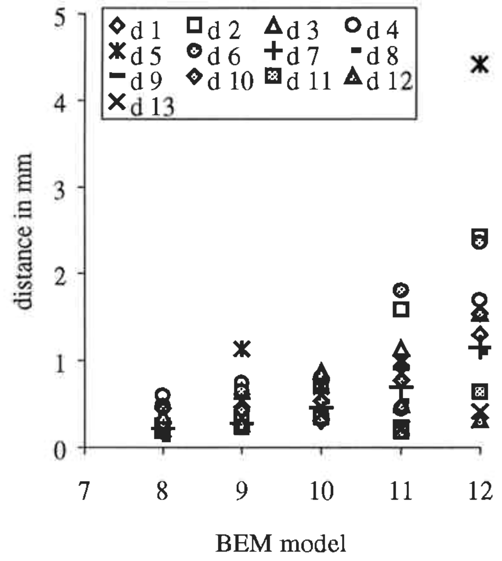

Fig. 3. Distance between original source location and inversely computed source location for all BEM models. BEM model 7 represents the reference model.

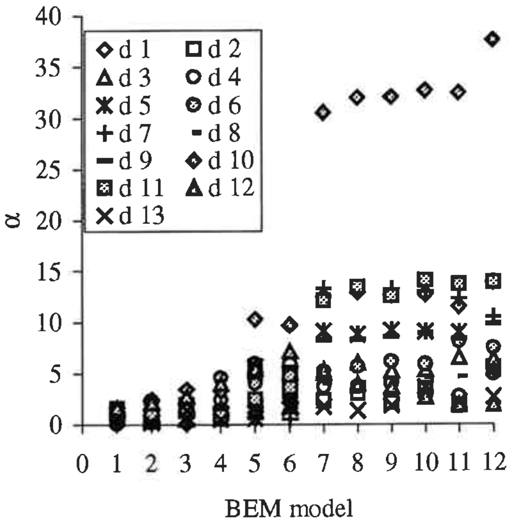

Fig. 5. Angle α between original dipole orientation and inversely computed dipole orientation for all BEM models.

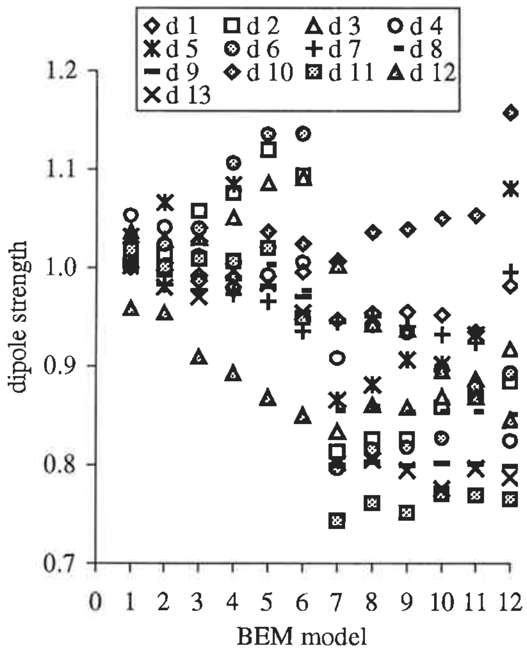

Fig. 4. Strength of inversely computed dipoles for all BEM models. BEM model 0 represents the reference model. The strength of the original dipole is equal to 1.0.

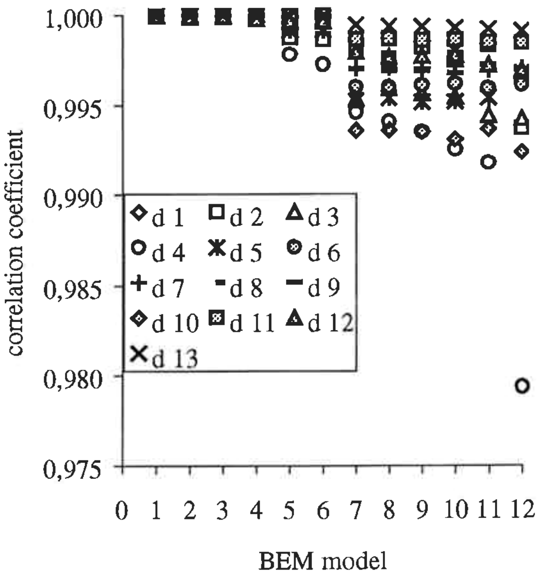

Fig. 6. Correlation coefficients between the reference field calculated with BEM model 0 and the fields calculated with the other BEM models.

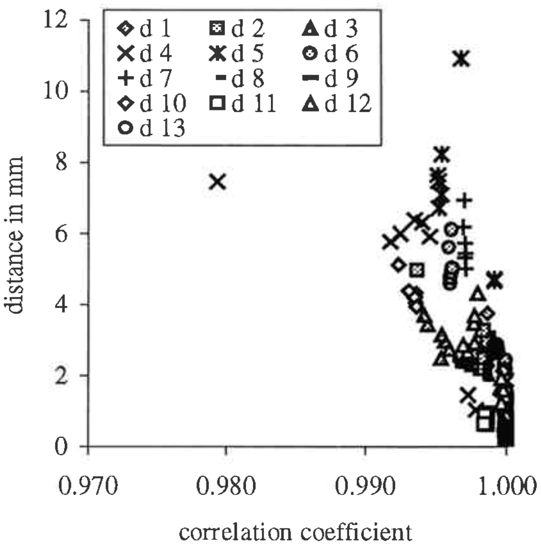

Fig. 7. Distance between original source location and inversely computed source location plotted against the correlation coefficient between the reference field calculated with BEM model 0 and the fields calculated with the other BEM models.

The influence of the BEM model modifications on the forward computation as measured by the linear correlation coefficient is depicted in Fig. 6. The correlation coefficient is computed between the simulated magnetic field map produced with the reference model (BEM model 0) and all other BEM models. The correlation is generally relatively high (above 0.99 except for one outlier: dipole 4, BEM model 12). The relation between the correlation coefficient and the source localization error is given in Fig. 7. Interestingly, a correlation coefficient above 0.99 can yield a localization error of up to 11 mm (Fig.7, dipole 5). Previous investigations [5,6] found correlation coefficient values above 0.999 indicate not more than 1-2 mm localization difference. In this study, we found that a value above 0.999 can indicate up to 3 mm localization difference.

For all computations performed in this study a standard desktop computer was sufficient. The inverse computations were divided in two parts. The first part was the setup and inversion of the kernel matrix (performed once for each model). The second part was the field computation for each source, which was performed up to several hundred times within one inverse computation. Typical values for the CPU time of a model setup were between a few minutes and 2-3 hours. The computation time for one inverse computation (without model setup) was in the order of a few milliseconds to a few seconds. The memory requirement for BEM model 0 was about 300 Mb.

IV. Conclusions

In conclusion, a triangle side length of 20 mm for the outer torso boundary, of 15 mm for the lungs, and of 8 mm for the ventricular blood masses are sufficient to reach an source localization error limit of approximately 2 mm. The correlation coefficient between magnetic field maps can serve only as a rough estimate for expected differences in source localization. The influence of BEM discretization on the reconstruction of extended cardiac sources [7] still needs to be quantified.

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