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Automatic generation of high resolution anisotropic FE models for EEG/MEG source reconstruction

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Abstract. EEG and MEG source reconstruction based on finite (FE) models represents the most sophisticated approach in the field of eletrophysiological neuroimaging. However, the generation of such models is time consuming and cumbersome. Therefore, we set up a toolchain of existing software packages in order to automatically generate models comprising five different tissue types and anisotropic conductivity information based on MR imaging data within less than 4 hours.

Keywords: EEG, MEG, source reconstruction, FEM, FE model

1. Introduction

The electromagnetic forward problem requires a volume conductor that models the electromagnetic properties of the investigated subject as accurate as possible. Typical types of forward models used in the literature in the field of EEG/MEG are spherical models, boundary element models, finite difference models and finite element models. The latter represents the modeling method with the highest level of sophistication, since it enables us to generate high resolution, inhomogeneous and irregulary-gridded models, which additionally allow the inclusion of anisotropy information. However, the use of such models suffers from large computational burden. Fortunately, the actual development in computer technology like multi-core hardware or GP-GPU (general purpose computing on graphic processing units) accelerated computing facilitates FEM calculations on standard computer hardware. Nevertheless, studies using this kind of modeling method are rare and show typically only one model [Wolters et al., 2006; Rullmann et al., 2010; Güllmar et al., 2010], which is mainly due to the time consuming and cumbersome generation of these models. Especially, the segmentation process is not always straightforward and easy to perform, since most segmentation tools cover only some compartments like (gray & white matter) or require a set of options, which have to be adapted for each subject individually.

Thus, the aim this study was to setup an easy to use tool chain of existing software tools, which is capable to automatically generate a segmented head volume comprising five different tissue layers (gray and white matter, csf, bone and soft tissue) as well as anisotropic conductivity information.

2. Material and Methods

2.1. MR imaging

All MR measurements were performed using a 3T MRI scanner (Tim Trio, Siemens Medical, Erlangen, Germany). T1w images were acquired using Magnetization Prepared Rapid Gradient Echo (MPRAGE) sequence with protocol parameters TE=3ms, TR=2300ms, TI=900ms, flip angle=9° and a full isotropic volume with a spatial resolution of $1mm^3$. T2w images were acquired using a SPACE – sampling perfection with application-optimized contrast using different flip angle evolutions – sequence, which is a variant of a conventional turbo spin echo sequence with refocusing pulse trains having variable flip angles. Sequence parameters were TE=355ms, TR=3000ms, flip angle=180 and a full isotropic volume with a spatial resolution of $(1.0\ mm)^3$. Diffusion tensor data were acquired using a TSRE-EPI sequence [Reese et al., 2003] with a spatial resolution of $2mm^3$, 30 different diffusion directions [Jones, 2004] and a b-value of 1000 s/mm^2. Calculation of the diffusion tensors was performed with *b*-matrices, which comprise the contributions of the imaging gradients and cross terms [Güllmar et al., 2005].

2.2. Segmentation of anatomical data sets

Segmentation of the brain

The registration of the different MR data sets was performed only on Dicom header information assuming no subject motion during the different scans. In case of the diffusion tensor data, we applied the rotational part of the transformation also to the diffusion weighting directions in order to obtain correct aligned tensors during tensor estimation. The segmentation of the brain was performed using the tool chain "recon-all" of the Freesurfer software package [Fischl et al., 2003]. In order to obtain segments for the gray and white matter as well as the ventricles, only the stages autorecon1 and autorecon2 have to be applied. These stages include B1 inhomogeneity correction, anisotropic filtering and co-registration to Talairach standard space, which allows hybrid atlas based segmentation of gray and white matter as well as the ventricles [Ségonne et al., 2004].

Segmentation of the skull and CSF

In order to obtain the most reliable segmentation of the skull we used both T1- and T2-weighted MRI data sets. Segmentation was performed using the brain extraction tool BET2 [Jenkinson et al., 2005]. The tool obtains the inner and outer skull layer by analyzing the intensity profiles perpendicular to the brain surface in a certain range (3 mm from inside to 60 mm outside). Since voxels representing the ventricles (left, right, third) were already identified during cortical segmentation, only CSF which surrounds the brain volume (between the pial surface and the skull layer) had to be segmented. We use the already defined skull layer, to created a mask by assigning each voxel within the inner skull layer that does not belong to gray matter, white matter or to the ventricles to CSF. This mask was then compared with the apparent diffusion coefficient (ADC) volume derived from DTI data. Voxels assigned to CSF by this procedure but having a low ADC value (lower than 2 $\mu m^2/ms$) were reassigned to soft tissue. These soft tissue segments occurred mostly at the supra-mid-sagittal part of the brain (c.f. Figure 1 d) and e)).

Remaining parts and model reduction

Remaining voxels located within the head surface but not assigned to gray or white matter, CSF or skull segments were defined as scalp or soft tissue layer. Inner air volumes were neglected, since the final model is cropped at the lower part of the brain to reduce the size of the model and to remove cavities (e.g., esophagus, air sinuses). This model reduction was achieved by generating a binary volume of the outer skull layer including everything inside. The binary volume was dilated iteratively to increase the object outline by 3 cm. Using an AND relation between the dilated object and a binary representation of the whole head object gave the final model outline. This last step truncates the segmented volume outside the skull, thus strongly reducing computation effort and memory demands. These image operations were performed using "*fslmaths",* which is part of the FSL software package. The segmentation procedure is shown step by steps in Figure 1.

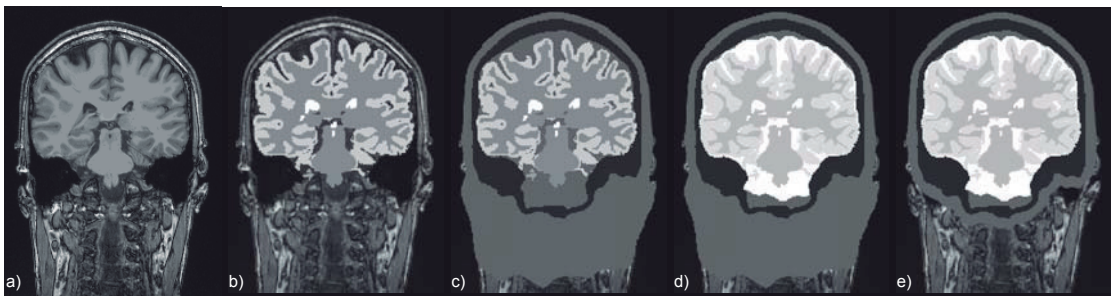

Figure 1. *Steps of the segmentation stepsSteps during the segmentation process using a T1w data set: a) coronal slice of an already filtered and normalized T1w image, b) segmentation of g/w matter as well as the ventricles, c) separation of the skull layer and the outer skin fills also the unsegmented tissues within the skull), d) classification of the CSF and dilatation of gray matter in the direction of white matter, e) lower part of the segment removed.*

Mesh preparation

For the finite element (FE) mesh generation hexahedral elements were used. This takes advantage of the cubic voxel structure inherent to MR images and strongly simplifies FE mesh generation for source analysis. Hexahedrization is performed by extracting the eight nodes of each mesh element directly from the corners of the voxels of the segmented volume. Since this procedure leads to a mesh, in which tissue interfaces are approximated by straight edges, right angles, and abrupt transitions, a node shift was applied using the algorithm proposed by Camacho et al. (1997). These steps were performed using *vgrid*, which part of the Simbio software environment [Simbio Development Group, 2011]. As a last step, electrical conductivity tensors were assigned to the white matter elements using the artificial anisotropy with volume constraint [Güllmar et al., 2010]. The amount of user interaction for the describe procedure is limited to pointing to the directory, which contains the Dicom files from MPRAGE and DTI MR data acquisition. The result of the automatic procedure is the FE model.In order to investigate the duration of the whole procedure, MR data sets of three healthy subjects were analyzed. As input for the tool chain only the Dicom data files derived from the MR scanner were used. After all automatic processing the time for model generation of all three subjects was averaged. With the current version 5.0 of *Freesurfer*, parts of the processing chain can take advantage of GP-GPU. In order to investigate the saving in time using GP-GPU, we performed the whole procedure without and with GPU (NVIDIA Tesla C1060 as well as C2050) on a workstation with 24GB working memory and Intel Xeon W3540@2.93GHz CPU running Ubuntu Linux 64bit.

3. Results

Figure 2 shows the final finite element model of one subject. It shows the five different tissue layers as well as the anisotropic conductivity tensor information placed in white matter elements. The typical size of the models was 3.5×10^6 elements. Table 1 shows the average time needed for model generation with and without using GP-GPU acceleration. The time saving using a NVIDIA Tesla C1060 was approximately 1h and 1h 30min using a NVIDIA Tesla C2050, which results in a speed up of 126% and 146%, respectively, compared to precessing without GP-GPU. Without GP-GPU acceleration the whole procedure took less than 5 hours.

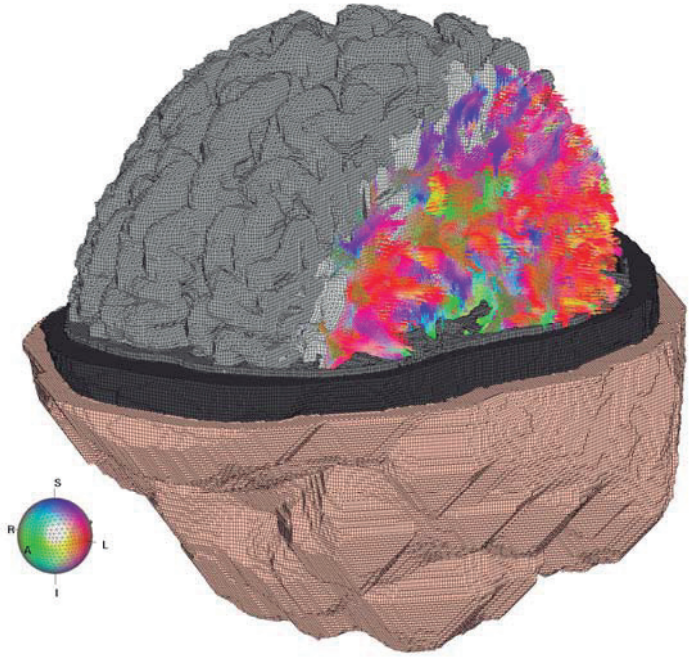

Figure 2. *Sample high resolution FE model comprising anisotropic conductivity information in white matter elements generated automatically. The skin, bone and csf layers are cropped in order to expose the gray and white matter segments. The elements of the latter comprise anisotropic conductivity tensors derived from MR diffusion tensor imaging.*

Table 1. *Duration for subcortical segmentation using freesurfer cortical reconstruction with and without using GPU acceleration. The mean values were obtained based on the processingof three different subjects.*

configuration	*recon-all duration*	*autorecon1 & autorecon2 duration*	*postprocessing and mesh generation*	*total duration*
without GPU	*10h 35min*	*4h 39min*		*4h 55min*
NVIDIA C1060	*7h 39min*	*3h 37min*	*16min*	*3h 53min*
NVIDIA C2050	*6h 3min*	*3h 6min*		*3h 22min*

4. Discussion

The proposed tool chain provides an easy and comfortable way to generate high-resolution anisotropic finite element models. Taking advantage of GPU computing can reduce the duration for model generation by 60 up to 90 minutes, depending of the type of GPU. But even without GPU acceleration the whole procedure can be performed within less than 5 hours on recent computer hardware. The tools used for model generation are freely available (Simbio requires registration). Mesh generation based on segmented datasets can also be performed using other mesh generators like *tetgen* [Hang, 2011], which is available without registration.

The procedure still has its limitations especially creating the skull layer. An additional MR contrast with proton-density weighting (PD) could help to improve discriminating soft and hard tissue within the skull layer.

Substituting the *freesurfer* tool (*recon-all*) with the *fast* (FMRIB's Automated Segmentation Tool) for the processing of gray/white matter as well as the ventricles might be an alternative. However, this tool requires an already extracted brain (e.g. using *BET*) and it may require adjustment of additional parameters in order to obtain optimal results. Also the use of SPM8 “New segment” might be an option. However, although SPM as well as FSL segmentation tools delivery a higher volume accuracy for segmentation procedure [Klauschen et al., 2009], *freeSurfer* was found to be more robust than FSL and SPM5 to changes in image quality, which in turn is more important in consideration of a full automatic segmentation procedure.

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