Predicting field distortions in the human brain using a susceptibility model of the head

Rebecca Sostheim[1], *Julian Maclaren*[1], *Frederik Testud*[1], *Maxim Zaitsev*[1]
[1]University Medical Center Freiburg, Freiburg, Baden Württemberg, Germany

Introduction Image artifacts due to motion during an MRI scan can be prevented using prospective motion correction [1]. However, the changes induced in the magnetic field within the measured object are not corrected for and thus can lead to geometric image distortion and signal loss. This problem occurs especially when measuring with gradient recalled echo (GRE) echo planar imaging (EPI) sequences. In order to take the field changes into account they either need to be measured in real time or need to be calculated [2]. In this work we identify the necessary components of a susceptibility model in order to predict field distortions depending on head position. We also compare two methods to generate this model.

Methods *Simulations:* All simulations were performed in MATLAB (The Mathworks, USA) using "Duke" of the "Virtual Family Model" [3] to generate the susceptibility distribution of a human. As a first step only the susceptibility values of soft tissue (χ = -9.2 ppm), bone (χ = -8.44ppm) and air (χ = 0.36 ppm) were used [4] and later only soft tissue and air. The starting point of the simulations was the entire body of this 2-component-Duke-model. Parts of the body were then neglected step by step. The corresponding computed fields for the brain were compared to the one using the entire body. RMSE was chosen to estimate the resulting error. The neglected part of the body was approximated by the field of a single magnetic dipole. The results from the simulations indicated the necessary components to model the field.

The main causes of the strong changes in the field are the air cavities in the head. Unfortunately these two components appear black on typical MRI images. Thus it is essential to use a model that accurately distinguishes between bone and air.

Atlas-based model: For the first model Duke was used as an anatomical atlas to give information about air. MR data were acquired with a 3D-GRE sequence (TE=3ms, TR=30ms, resolution=2mm^3, FOV=256mm^3, flip angle=10°) at 3T (Tim TRIO, Siemens, Germany). Duke was then matched to the MRI data using point based registration in MITK (DKFZ, Heidelberg, Germany). The registered sets of data were then merged into one model in MATLAB by using information about soft tissue from the MRI data and information about bone and air from Duke.

UTE-based model: For the second model measurements were done using a 3D ultra-short TE (UTE) sequence with two contrasts (TE_1=0.07ms, TE_2=3.76ms, resolution=265mm^3, FOV=256mm^3, flip angle =10°) [5]. These sequences often show a better contrast of bone and air. A threshold based on the intensity of the data was chosen for each contrast to separate air from all other tissues. The data sets were combined into one model. Magnetic fields for both models were calculated.

Field maps of a volunteer's head were acquired using a double echo GRE sequence (TE_1=3ms, TE_2=5,46ms, TR=30ms, FOV=256mm^3, flip angle=10°).

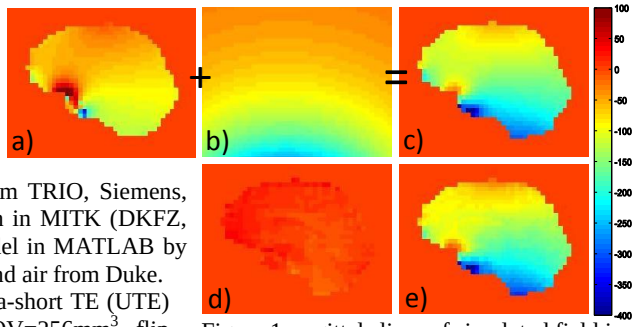

Figure 1: sagittal slices of simulated field in Hz: a) air-soft tissue from jaw upwards, b) fitted dipole field, c) 1a) plus 1b), e) ground truth, d) difference between 1c) and 1e)

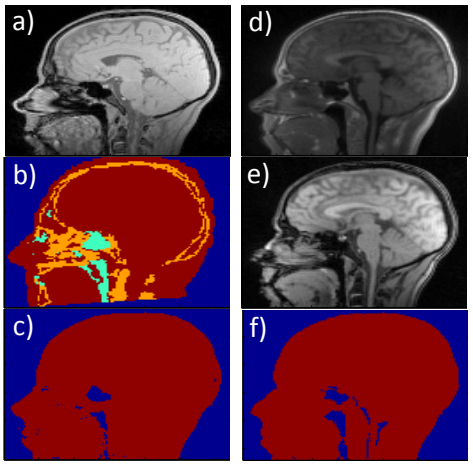

Figure 2: sagittal slices of : a) MR data from 3D-GRE, b) Duke, c) atlas-base model, d) UTE contrast 1, e) UTE contrast 2, f) UTE model

Results and Discussion Figure 1 shows sagittal slices of the simulated magnetic field of the brain before (e) and after (c) the simplification. Comparing 1c) and 1e) it can be seen that the difference in the field distribution is very small (Fig. 1e): RMSE = 6.43 Hz. Hence, it is valid to use a theoretical model that consists of only two susceptibility values (air and water) and includes only the head (from the jaw upwards). The rest of the body can be approximated through a magnetic dipole field (Fig. 1b), placed 21 cm below the circle of Willis.

Figure 2 shows the steps leading to the models. Figure 2c) shows the atlas-based model after the registration of 2a) and 2b). Figure 2f) shows the UTE-based model that was generated by thresholding and combining the UTE data in 2d) and 2e).

Figure 3 shows sagittal slices of the measured field map (a), the field map computed with the atlas-based model (c) and the field map calculated for the UTE model (c). The results show that it is possible to generate a model with the methods described above. The simulated fields for both models look very similar. It is therefore a great advangtage to know that it is possible to use UTE data because it makes generating the model much faster and eliminates the manual registration step. The simulated fields are almost a perfect match to the real field map in the back of the brain but show differences in the front. The problems here seem to be related to very small air cavities in the sinuses for which the spatial resolution of the two models is apparently insufficient. Susceptibility mapping approaches [6] may be used to recover the source of the B_0 variation in the frontal areas.

Conclusion: In this work we have shown that a simple two-component susceptibility model can be used to estimate the field in the human brain. A UTE-based model is a promising alternative to an atlas-based model, as no manual registration steps are required. With such a model it should be possible to predict field distortions for a new head position after motion without having to acquire a field map.

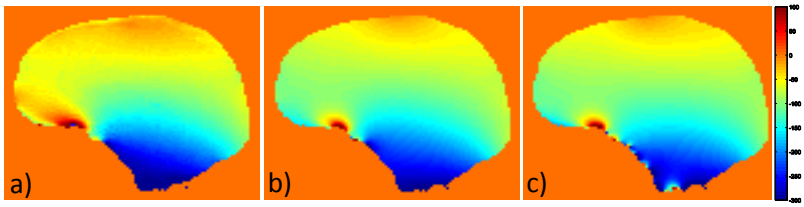

Figure 3: field maps in Hz: a) measured b) for Atlas-based model, c) for UTE-model

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