In-Vivo Human Brain Inhomogeneity Estimation using Susceptibility Registration and Rapid Field Simulation

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Introduction Differences in magnetic susceptibility can induce significant perturbations in static fields utilized for magnetic resonance. While susceptibility differences can be exploited as contrast mechanisms (i.e. the origins of the BOLD signal in fMRI), they typically introduce unwanted inhomogeneity in the static field. Knowledge of these induction fields is thus an important issue for many magnetic resonance applications. Field inhomogeneity is commonly determined with image-based field map acquisitions, which can become unacceptably long depending on desired resolution and precision. Recently, induction field simulations have been introduced which exploit the small susceptibilities of materials compatible with magnetic resonance studies [1] [2] [3]. Two of these methods [1][2] result in solutions of the induction field through Fourier transforms of the susceptibility distribution. Field solutions of this nature can be calculated in a matter of seconds on a standard workstation at high resolution and therefore present a time-efficient alternative to field map acquisitions.

Utilization of field simulations for such purposes requires robust estimation of sample susceptibility distributions. The large-scale nature of induction fields in the human brain primarily results from air-tissue interfaces. This is because the susceptibility difference between air and water is approximately 10 ppm while the susceptibility of all biological materials in the brain differ from that of water by at most a few tenths of a ppm. To a first approximation, a susceptibility model for macroscopic trends in field inhomogeneity therefore requires only a two-compartment distribution of air and water. However, MRI used for susceptibility modeling will contain signal voids consisting of both bone and air. A means of quickly decoupling the air and bone from these signal voids is required to utilize MRI for rapid susceptibility modeling. Here, we present a method whereby registration of MRI to an existing model of the sinus and auditory cavities can be used to build an accurate estimate of the susceptibility distribution. Fast field simulations of these susceptibility distributions then provide noiseless estimations of inhomogeneity within the brain that can be used for shimming and post-acquisition correction of geometric distortion.

Methods The field simulation method of Salomir, Senneville, and Moonen [1] was implemented in this investigation. In this approach, a small susceptibility approximation is made in Laplace's equation for the magnetic scalar potential. A Fourier transformation solution to the approximated Laplace equation results in an analytic approximation to the field given by $B_z(\mathbf{r}) = B_0\ FT^{-1}\{(k_z^2/k^2 - 1/3)FT[\chi(\mathbf{r})]\}$, where $\chi(\mathbf{r})$ is the 3D susceptibility distribution, FT represents a 3D Fourier transform, FT^{-1} represents the inverse transform, $\mathbf{k}$ is the transform space coordinate, and the field measurement is assumed to be in a spherical Lorentzian susceptibility cavity.

Simulations were implemented on a 2.8 GHz Dell Precision Workstation with 512 Mb of RAM using vectorized code written in Matlab. For a resolution of 96x96x64 the simulation required less than 3 seconds. Comparison of the field simulation with analytic predictions of susceptibility distributions utilizing spheres and cylinders showed agreement with results previously reported.

Identification of air cavities was performed using nonlinear intensity-based registration [4] and a segmented human head model [5]. The head model was simplified to a binary segmentation of air and biological matter. Registration of MRI source data from the binary susceptibility phantom to acquired images provided models approximating the susceptibility distribution within the subject. Construction of this susceptibility model required less than 90 seconds of processing time for 6 iterations of the registration algorithm. The total time for an inhomogeneity estimation including image acquisition, susceptibility modeling, and field simulation required less than 3 minutes for a resolution of 96x96x64 pixels.

Accurate simulation of total field inhomogeneity also requires determination of the subject-free magnet inhomogeneity. This was accomplished through empirical field mapping of a spherical water phantom. While the theoretical induction field inside a spherical cavity is spatially invariant, any fabricated spherical phantom has imperfections. Therefore, the induction field from the imperfect spherical phantom was simulated and removed from the acquired field map.

Images for susceptibility modeling and *in-vivo* magnetic field maps were acquired on a 4.0 T Magnex magnet interfaced to a Bruker Avance spectrometer. Reception and transmission were performed with a Bruker TEM coil. Field maps were acquired with an asymmetric gradient echo sequence over 64 0.3 mm slices with 96 x 96 pixels over 25.6 x 25.6 cm. As previously described, a binary air-water susceptibility model was estimated from the MRI. The induced field was then simulated at the same spatial resolution and compared to the measured field map with the subject-free background subtracted.

Results and Discussion Figure 1A displays an axial anatomic slice from the 3D acquisition located slightly above the sinus cavities. Line segments indicate traces of the simulated and background-corrected measured induction fields for sagittal and coronal directions. Figures 1B and 1C present these traces with the induced field offset expressed as parts per million (ppm) of the 4 Tesla static field..

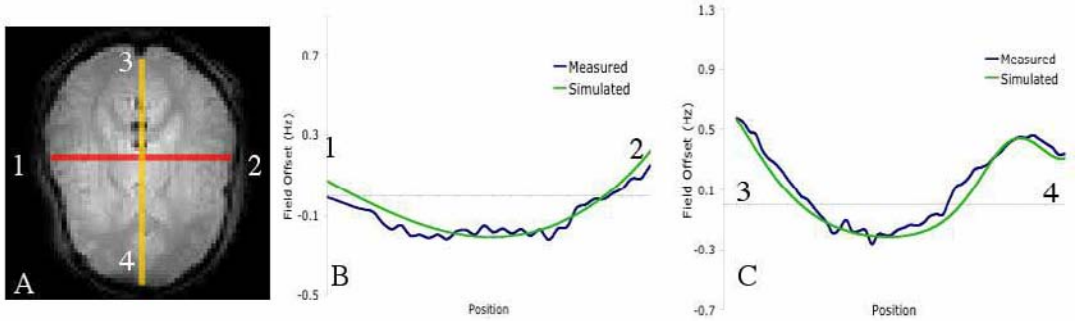

Figure 1: A) MRI of slice located slightly above sinus cavities illustrating sagittal (red) and coronal (yellow) inhomogeneity profiles, B) sagittal trace (3-4) through 3D simulated field and experimentally measured inhomogeneity, and C) coronal (1-2) trace.

The simulation shows good agreement with the general trend of the experimental measurement. The observed agreement is more than acceptable for use in large volume shim optimization. For inhomogeneity estimation utilized in post-processing, the number of iterations and sophistication of the nonlinear registration can be increased to more accurately localize air cavities. Field simulations using these accurate susceptibility models can then more successfully model higher order spatial inhomogeneity. Such higher order effects are responsible for the significant geometric distortions found in EPI data. Post-acquisition distortion correction methods could utilize these higher-order field estimations to improve spatial quality of EPI data.

In conclusion, we have presented a fast, accurate B_0 inhomogeneity estimation method that can be utilized for shim optimization or post-acquisition corrections. Further improvements can be made through increasing the accuracy of our air cavity model. The phantom used in this study identifies hundreds of anatomic compartments at the expense of sacrificing the accuracy of individual compartments. Since we are concerned only with air compartments, through closer analysis of the source MRI and CT data we plan to increase air cavity localization in the phantom.

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