Diffusion-Guided Quantitative Susceptibility Mapping

Amanda C. L. Ng1,2, David K. Wright3,4, Parnesh Raniga2,3, Stephen Moore4, Gary E. Fegan2, and Leigh A. Johnston5

1Dept of Electrical & Electronic Engineering, The University of Melbourne, Melbourne, VIC, Australia, 2Monash Biomedical Imaging, Monash University, Melbourne, VIC, Australia, 3Centre for Neuroscience, The University of Melbourne, Melbourne, VIC, Australia, 4Centre for Neuroengineering, School of Computing and Electrical Engineering, The University of Western Australia, Perth, WA, Australia, 5CSIRO Preventive Health National Research Flagship ICT, Herston, QLD, Australia

Target audience Researchers investigating magnetic resonance phase imaging and susceptibility mapping.

Purpose Quantitative susceptibility mapping (QSM) aims to derive reliable estimates of the magnetic susceptibility of voxels from phase data arising from 3D gradient-echo MRI acquisitions. Current approaches model every voxel as a sphere5, which however research has demonstrated that white matter voxels are better modelled as cylinders. We propose a new approach to QSM that uses diffusion-weighted MRI to guide geometric model selection in each voxel. We demonstrate that the diffusion-guided QSM method is more accurate and robust than conventional methods.

Methods

Theory $\Delta B$ is calculated from $T_2^*$ GRE phase data according to $\Delta B = -\gamma TE \phi^{-1}$, where $\gamma$ is the gyromagnetic ratio of water, $TE$ is echo time and $\phi$ is phase. The susceptibility map, $\Delta \chi$, is related to $\Delta B$ according to $\Delta B(r) = \sum F(r', r-r') \Delta \chi$, where $F(r', r-r') = F_1(r-r')$ for spherically modeled voxels and $F(r', r-r') = F_0(r-r')$ for cylindrically modeled voxels. The spherical kernel function is given by

$$F_0(r) = \left\{ \begin{array}{l l} \frac{3}{4 \pi} \frac{1}{r^3} \phi, & r > 0 \\ 0, & r = 0 \end{array} \right.$$ (1)

where $r = \rho \theta$, $\theta$ is the angle between $r$ and $B_0$-field direction, $\alpha$ and $\beta$ are the spherical coordinates $\alpha$ is the angle between the projection of $r$ and $x$ onto the plane normal to $c$, $\beta$ is the angle between $c$ and $z$. The proportionality function

$$P(r) = \frac{2(2-r^2)^{\alpha}}{4\pi^2} (\alpha + r^2 c), \alpha = 0.47$$ (2)

facilitates the discretisation of the analytical kernel, where $\alpha$ was determined computationally to optimise the discrete approximation of the continuous kernel. Fractional anisotropy (FA) and primary eigenvector (V1) maps are calculated from DWI data. Voxels with $FA \leq 0.2$ are modelled as spheres, while voxels with $FA \geq 0.2$ are modelled as cylinders whose axes are defined by $V1$. $\Delta \chi$ was solved by minimising $\| Ax - b \|_2 + (1 - \kappa) \| Lx \|_2$, where $Ax - b$ is the matrix-vector representation of $\sum F(r', r-r') \Delta \chi = \Delta B(r)$ and $L$ is a second-order derivative.

Simulation Data The dQSM method was applied to a numerical phantom comprising 4 cylinders and 4 spheres paired with matching $\Delta \chi$ of 1e-7, 2e-7, 3e-7 and 4e-7. The cylinder axes were oriented at 90° to the $B_0$ field. The matrix size was 50×125×75 and radii of cylinders and spheres were 5 voxels. Experimental Data Ex-vivo mouse brain $T_2^*$ GRE (3D EPI, TR=1000ms, TE=100ms, FA = 30°) and DWI (TR=2500ms, TE=65ms, shots=2, 5=3ms, $\Delta t=14ms$, 46 drrs, b=1700 s/mm$^2$) data were acquired in a single scan session on a 4.7T Bruker with MTEX=192×168×96, voxel size=0.1×0.1×0.1mm$^3$. The GRE magnitude and DWI $B_0$ images were coregistered using FSL FLIRT. The GRE phase data was unwrapped with $\Phi^{UN}$ and filtered with SDF². Computation The $\Delta \chi$ data maps were calculated on an IBM BlueGene/Q, taking 16 hours to complete the experimental data maps on 4096 cores. The Landweber iteration was used to compute the minimisations. A lower threshold of 10$^{-5}$ was applied to the kernel values. $\kappa$ was set to 0.75.

Comparison method MEDI-derived susceptibility maps were computed for comparison. The $\lambda$ parameter was set to 0.1 based on qualitative analysis of artefact removal and smoothing.

Results

The numerical phantom results (Fig. 1) demonstrate accurate computation of the susceptibility values of the cylinders and spheres for the dQSM method. In contrast, the MEDI method under-estimated the susceptibility values, and computed different values for cylinder-sphere pairs with equal susceptibility values. The ex-vivo mouse results (Fig. 2) demonstrate more uniform susceptibility values in the white matter of the corpus callosum (white arrows). Structure visible in the magnitude image (square) is not visible in the MEDI map, but does appear in the dQSM map. The MEDI map appears noisier than the magnitude and dQSM map. Arrows indicate white matter correctly resolved by dQSM.

Discussion

dQSM has demonstrated enhanced ability to resolve $\Delta \chi$, particularly in the white matter, where cylindrical geometries dominate. While the $\Delta \chi$ values derived by MEDI are known to scale with the regularisation weighting$^2$, dQSM shows invariance and accuracy of the estimated $\Delta \chi$ values. The main drawback of the proof-of-principle dQSM method is the high computational costs. Current QSM methods involve only a spherical kernel can invoke the convolution theorem, thereby substantially reducing computation time to $\text{Nlog}(N)$ by employing the Fourier transform. Since dQSM involves spatially dependent kernels, the convolution theorem is no longer applicable and computation time is high at N$^3$. Current work is needed to increase the efficiency of the dQSM approach.

Conclusions

We have demonstrated that using diffusion weighted MRI to guide the selection of cylindrically modelled voxels increases the accuracy of estimated susceptibility values. Our proof-of-concept dQSM method, while computationally expensive, provides a first step beyond the Lorentz sphere model assumption.

References