Compensating for Within-Voxel Susceptibility Gradients in BOLD fMRI

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Introduction: Blood oxygenation level dependent (BOLD) functional MRI relies on contrast due to susceptibility differences between oxygenated and deoxygenated hemoglobin [1]. To sensitize acquisitions to this BOLD effect, studies are performed at high field strength using single-shot acquisitions. But these parameters also make the acquisitions sensitive to susceptibility differences between adjacent structures in the brain, leading to susceptibility artifacts in the images. The conjugate phase method has been the dominant algorithm used to compensate for these susceptibility effects [2,3]. Recently, an iterative reconstruction method has been presented that models off-resonance phase during the course of the readout and reconstructs the corrected image via an inverse problem approach [4]. However, neither of these correction methods addresses signal loss due to gradients in the field inhomogeneity distribution within a voxel. These gradients cause phase dispersion and cancellation of spins within a voxel. In this work, we propose two methods for correcting for signal dropout due to within-voxel susceptibility gradients; one method uses a piece-wise linear model for the local resonant frequency within each voxel, the other method uses an oversampled field map and a corresponding piecewise constant model within each voxel.

Theory: The signal equation for MRI is given by: $s(t) = \int f(\mathbf{r}) \exp(-i\omega(\mathbf{r})t) \exp(-i2\pi \mathbf{k}(t)\cdot\mathbf{r}) d\mathbf{r}$, where s(t) is the baseband signal at time t, f(**r**) is the object to be imaged, ω(**r**) is the value of the local off-resonant frequency at location **r**, and **k**(t) is the k-space trajectory. In [4], both the object and field map were expanded in terms of a limited number of rectangular basis functions, i.e., voxel indicator functions. That model fails to account for within-voxel variations in resonant frequency, so within-voxel field gradients still result in apparent signal loss. In this work, we expand the field map with a piece-wise linear basis function, having both an off-resonance value and a slope in x-, y- and z-directions for each pixel location as:

$$\omega(x,y,z) = \sum_{n=1}^{N} \left(\omega_n + X_n \frac{x-x_n}{\Delta x} + Y_n \frac{y-y_n}{\Delta y} + Z_n \frac{z-z_n}{\Delta z} \right) \mathrm{rect}\left(\frac{x-x_n}{\Delta x}, \frac{y-y_n}{\Delta y}, \frac{z-z_n}{\Delta z} \right)$$

, where X_n, Y_n, Z_n are the x,y,z-gradients across the voxel at location $\mathbf{r}_n=(x_n,y_n,z_n)$ and $\Delta x,\Delta y,\Delta z$ is the size of the voxel. Alternatively, we could use a higher resolution field map in our system model and enforce the original resolution through an upsampling/downsampling operation. This would inherently allow us to reduce within voxel dephasing through the use of smaller voxels.

Methods: A simulation study was performed using a 256x256 matrix size simulation object and simulating a spiral acquisition for a 64x64 matrix size. Three 1mm slices were combined to form a 3mm thick slice. Reconstruction was performed using the iterative method of [4] (field corrected), including first order gradients (FC with gradients), and a two-times oversampled approach. A phantom study was performed using a susceptibility phantom that consisted of a water-filled cylinder with the inferior ventral semi-cylinder section filled with air. A field map was collected at a 64x64 matrix size for 1.6mm slices. The data to be reconstructed was collected at a 64x64 matrix size with 4.8mm slices. An undistorted image was acquired using a 4-shot spiral sequence.

Results:

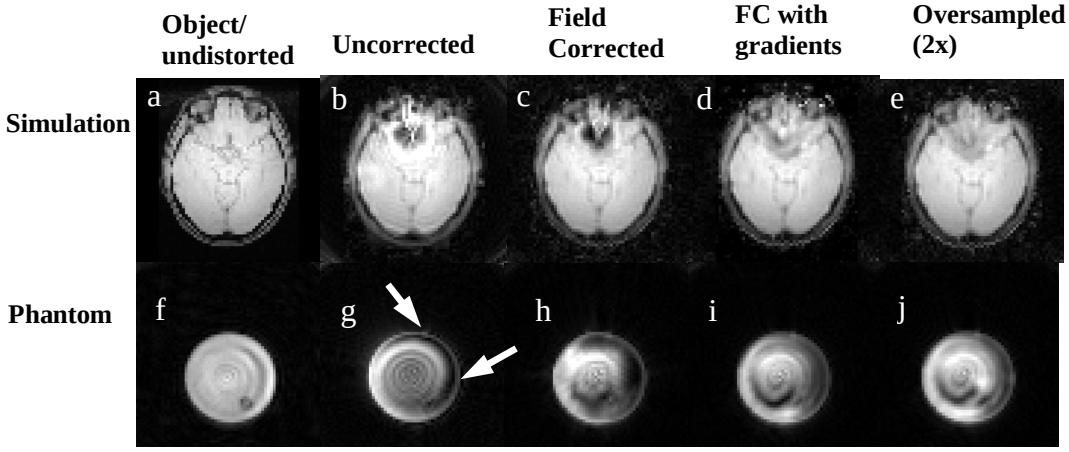

Figure 1: Results of simulation (a-e) and phantom (f-j) studies. Panels (a,f) represent the undistorted object, (b,g) are the uncorrected images, (c,h) are the field corrected images using piecewise constant basis functions for the field map, (d,i) are the reconstructions with a piecewise linear basis, and (e,j) are the two-times oversampled field map reconstructions.

Discussion and Conclusion:

Including a model of the within-voxel field inhomogeneity distribution can allow us to recover some signal from regions plagued by susceptibility-induced signal drop out. We have presented two approaches, a model that includes first order gradients and an oversampled field map technique.

References: [1] Ogawa, *et al.* Proc. Natl Acad Sci., 89:5951-5955, 1992. [2] Schomberg. IEEE Trans Med Im. 18:481-495, 1999. [3] Noll, *et al.* IEEE Trans Med Im. 10:629-637, 1991. [4] Sutton, *et al.* IEEE Trans Med Im. 22:178-188, 2003.

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