Inherent Motion Correction for Multi-Shot Spiral Diffusion Tensor Imaging

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Introduction

Multi-shot spiral imaging is a promising alternative to echo-planar imaging for high-resolution diffusion tensor imaging (DTI) (1–5). However, subject motion in the presence of diffusion-weighting gradients causes phase inconsistencies among different shots, resulting in signal loss and aliasing artifacts in the reconstructed images. Such artifacts can be reduced by using a variable-density spiral trajectory (1–4) or a navigator echo (5), however at the cost of a longer scan time. In contrast, an iterative phase correction method was recently proposed to inherently correct for these motion-induced phase errors with no scan time penalty (6) and proof-of-concept simulations were presented for a 2-shot spiral acquisition. Here, we further extend this novel method to M-shot spiral acquisitions (with $M \geq 2$) and demonstrate its effectiveness in real DTI experiments.

Methods

We propose a new theoretical framework based on the following equation: **a** = **E u** [1]. For each pixel (x_0, y_0) of an $N \times N$ image, **a** is an $M \times 1$ vector containing the pixel values from the M aliased images reconstructed from each shot separately (by zero-filling the missing k-space data from the other shots) (Fig. 1). **E** is an $M \times N^2$ matrix whose rows contain $N \times N$ subsets of the point spread function (computed from the k-space trajectory) as well as $\exp[-i\phi_m(x_0, y_0)]$, where ϕ_m is the motion-induced phase error between the m^{th} shot and the first shot (for $m \geq 2$). Finally, **u** is an $N^2 \times 1$ vector whose $(x_0, y_0)^{\text{th}}$ element contains the pixel value from the unaliased image to be reconstructed.

As such, if ϕ_m is known (e.g., from a navigator echo), the unaliased image can be determined by solving Eq. [1] for each pixel. However, since ϕ_m is generally unknown, we use an iterative phase correction method, which consists in reconstructing a series of images with different ϕ_m values and choosing the image with the least aliasing. The latter is chosen as the image that has the lowest background energy, which is defined as the sum of the signal intensity in all background pixels. The background is determined from the baseline (b = 0) DTI image, which is not affected by motion-induced phase errors.

Here, we assume that ϕ_m is spatially linear, i.e., $\phi_m(x, y) = \phi_{0,m} + g_{x,m}\, x + g_{y,m}\, y$, and step through different $\phi_{0,m}$, $g_{x,m}$, and $g_{y,m}$ values, where $\phi_{0,m}$ is a global phase offset due to translational motion, whereas $g_{x,m}$ and $g_{y,m}$ are linear phase gradients due to rotational motion (7). This model is thus sufficient to correct for rigid-body motion, but can also be extended to correct for nonlinear phase errors caused by nonrigid motion.

Since reconstructing images with all possible ϕ_m values would be time-consuming, we use the following strategies to drastically reduce the computation time. The phase optimization is performed: i) on low-resolution images reconstructed from the central k-space, ii) iteratively by reducing the range and step size for the $\phi_{0,m}$, $g_{x,m}$, and $g_{y,m}$ values at each iteration, and iii) by solving Eq. [1] only in the background pixels, since only those contribute to the background energy. Once the optimal ϕ_m have been determined, the final image is then reconstructed at full resolution and in all pixels with these values.

We studied healthy volunteers on a 3T GE scanner with a 4-shot constant-density spiral DTI sequence, TR = 5 s, TE = 56 ms, FOV = 24 cm, matrix size = 160×160, slice thickness = 3 mm, b-factor = 1000 s/mm^2, and 6 diffusion-weighting directions. Pulsatile motion artifacts were minimized with cardiac gating and rigid-body motion was corrected with the proposed method. The phase optimization was performed with a resolution of 12×12, five iterations, and an initial step size of π for the $\phi_{0,m}$, $g_{x,m}$, and $g_{y,m}$ values.

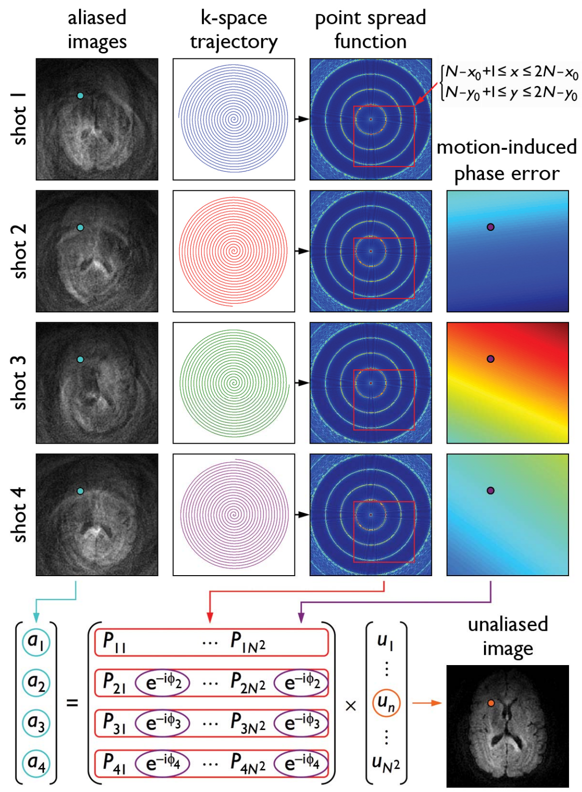

Fig. 1: Schematic diagram of Eq. [1] for a 4-shot spiral DTI acquisition.

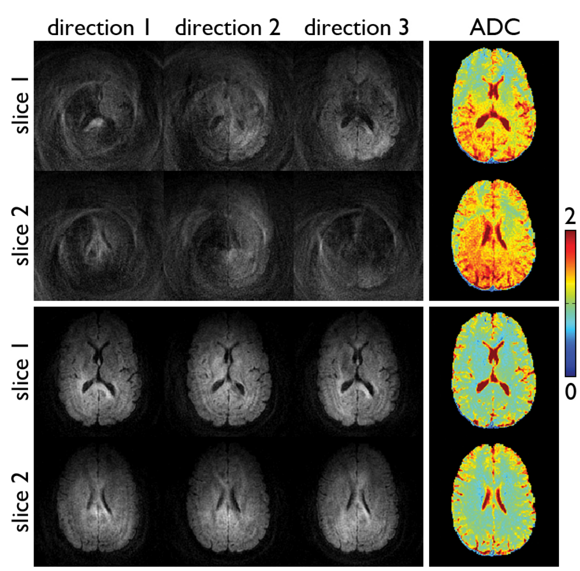

Fig. 2: Uncorrected (top) and corrected (bottom) diffusion-weighted images for three representative diffusion-weighting directions and resulting ADC maps (in 10^{-3} mm^2/s).

Results and Discussion

Representative results are shown in Fig. 2. Even though the subjects were instructed to remain still and their head was restrained with padding, residual motion still causes severe signal loss and aliasing artifacts in the diffusion-weighted images as well as subsequent errors in DTI metrics such as the apparent diffusion coefficient (ADC) or fractional anisotropy The proposed method can correct for both types of artifacts and significantly improve the image quality. The computation time is about 3 h, but can be further reduced by using faster optimization algorithms and/or parallel computing.

These preliminary results demonstrate that the proposed iterative phase correction method can inherently correct for the spatially linear phase errors, as well as the resulting signal loss and aliasing artifacts, caused by rigid-body motion in multi-shot spiral DTI, without requiring a variable-density spiral trajectory or a navigator echo, and hence without increasing the scan time. Additional work is currently underway to further improve and validate this method, which should be particularly useful for high-resolution DTI studies of pediatric and patient populations who cannot tolerate long scan times.

References: 1. Liu MRM 2004;52:1388, 2. Van IEEE TMI 2009;28:1770, 3. Karampinos MRM 2009;62:1007, 4. Frank NeuroImage 2010;49:1510, 5. Van IEEE TMI 2011:in press, 6. Truong ISMRM 2011;19:4594, 7. Anderson MRM 1994;32:379. This work was in part supported by NIH grant EB12586.