Diffusion Weighted MRI of Moving Subjects based on Motion-Induced Random Oversampling

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Purpose

Recent work on motion correction in diffusion-weighted magnetic resonance imaging (DW-MRI or DWI) has been quite promising (). Most techniques rely on real-time motion detection to update field of view and to repeat motion-corrupted scans[1,2]. However, this strategy is not always effective, for example in applications where the subjects move continuously and randomly. Moreover, real-time detection of motion is not always straightforward, for example it is difficult to detect small movements of the fetal head in fetal DWI. DWI is specifically demanding on gradients and is noisy; it often provokes fetal and neonatal motion and may wake neonates[3]. Neonates may startle and move with the vibration and acoustic noise from a DWI sequence. There has been invaluable recent work on motion-robust DWI of moving subjects[4,5]. The essence of these methods is fast DWI slice acquisition, slice-by-slice motion correction, and scattered data interpolation (SDI) to combine motion-corrected slices. As compared to these methods, we propose a novel approach that corrects for motion at the slice level and upon alignment of images and rejection of motion-corrupted data, calculates the diffusion components at the voxel level. Sufficient coverage of the k-space, image space, and q-space is achieved in our approach through random oversampling. Uncontrolled subject motion imposes random sampling in image space and q-space. Oversampling, on the other hand, is performed purposely in our approach to obtain sufficient coverage of the image space and q-space.

Methods

A DWI study involves one or a few reference images without diffusion sensitivity (b=0 image) denoted by S_0, and a set of diffusion-sensitized images, denoted by S_i, with non-collinear gradient directions G_i. The relation between the diffusion-sensitized signal at point x and the gradients is described by the Stejskal-Tanner equation (where γ is the gyromagnetic ratio, δ is the gradient duration, and Δ is the time between two pulses): Eq.1:

$$ln\left(\frac{S_i(\boldsymbol{x})}{S_0(\boldsymbol{x})}\right) = -\gamma^2 \mathbf{D}_i \delta^2 \left(\Delta - \frac{\delta}{3}\right) \|\mathbf{G}_i\|$$

If imaging is perfect, exact correspondence exists between S_i and S_0; however, in the presence of motion, voxels with the same image coordinates do not correspond to the same physical points. This invalidates Equation 1 and induces error. The first stage of our algorithm is slice motion correction, which is performed through the registration algorithm shown in Fig. 1. The outcome will be a 6-parameter rigid transformation for each slice, which maps the slice to the S_0 volume. It is crucial to update the gradient directions corresponding to each slice with these parameters. Motion-corrupted and mis-registered slices are detected as outliers based on registration metrics and are eliminated. The corrected DWI data is then mapped to the high-resolution lattice of the target image.

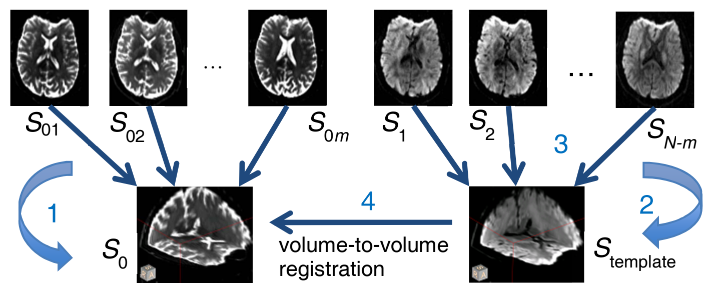

Fig. 1: Our DWI slice registration algorithm has four stages: 1) reconstruct a S_0 volume, 2) construct a DWI template ($S_{template}$), 3) register all slices to $S_{template}$, and 4) register $S_{template}$ to S_0.

Data

DW-MRI was obtained from 3 volunteers on a Siemens Trio 3T scanner with a 32-channel head coil. DWI was performed with isotropic resolution of 2mm, with 30 gradient directions, and 5 b=0 scans. Each experiment involved one scan without motion (used as reference) and two scans with motion.

Results

First, we evaluated the accuracy of our DWI slice-to-volume registration with a set of experiments under controlled two-position head condition. In these experiments volume-to-volume registration provided ground truth to evaluate slice-to-volume registration. The results showed median absolute error of less than 1mm in translation and less than 2 degrees in rotation parameters. Although registration failed for a number of slices, those slices were detected and rejected as outliers through the analysis of registration metrics. Fig. 2 compares the color FA of DWI analysis for the reference scan, and for DWI before and after motion correction with gradient update.

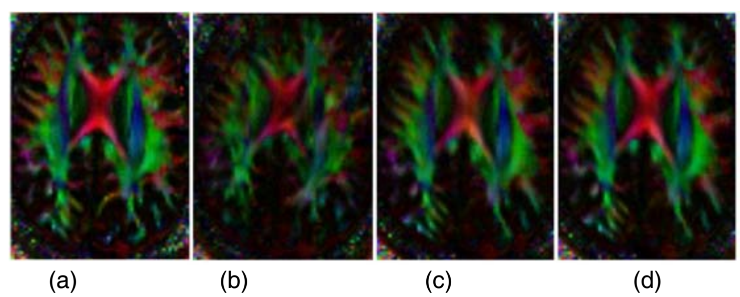

Fig. 2: Color fractional anisotropy (color FA) in a volunteer DWI study: (a) reference color FA obtained from an experiment without motion, (b) color FA of an experiment with motion, (c) color FA after motion correction without gradient update, and (d) color FA after motion correction and reconstruction with gradient update. The Peak-Signal-to-Noise Ratio (PSNR) between the reference and (b), (c), and (d) is 21.84, 27.20, and 29.77, respectively, which indicates that the motion is compensated and signal is restored in (d).

Discussion

At least 6 non-collinear gradient directions are needed to calculate a rank 2-tensor model, but typically more than 15 directions are used. Our approach relies on oversampling in q-space and image space to compensate for the effects of subject motion. This is particularly useful for subjects that move constantly and irregularly during scans. Future work aims at DWI of fetuses and non-sedated neonates.

References

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