A Framework for Evaluating the Performance of EPI Distortion Correction Strategies in Diffusion Tensor MRI

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Introduction

Diffusion weighted images (DWIs) used for Diffusion Tensor (DT) calculation are commonly acquired with Echo-planar imaging (EPI). B_0 inhomogeneities affect EPI by producing spatially nonlinear image distortions. These distortions degrade the anatomical accuracy of DT-MRI maps and may increase their variability in both longitudinal and cross-sectional clinical studies. Several strategies have been proposed to correct EPI distortions, such as B_0 field mapping [1-3], point spread function (PSF) mapping [4], and image registration [5,6]. All these methods remain virtually unused in the context of DT-MRI, in part because there are no objective tests for establishing if the improvement in the quality of the computed DT-MRI data justifies the additional scan time and/or computational complexity that these methods require. In this study, we propose a test to evaluate and compare the performance of different EPI distortion correction methods in improving the quality of DT-MRI results. We then apply our test to evaluate the ability to improve DT-MRI of a simple image-registration-based distortion correction scheme.

Methods

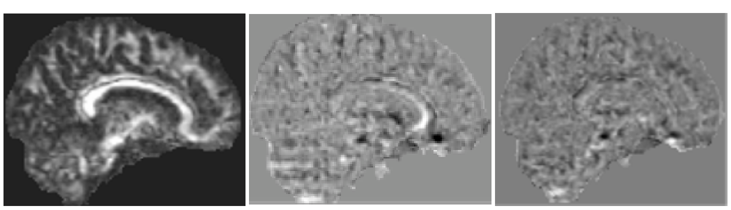
Methodological framework. The regional variability of DT-MRI data for a given experimental design in a given subject can be measured directly by computing the variance of tensor derived quantities on a voxel by voxel basis from tensors computed from co-registered replicate DWI datasets. Given that EPI distortions occur in the phase encode direction, one can obtain tensors that are differently corrupted by the EPI distortions by acquiring DWI datasets that differ only in the direction of phase encoding. The local variance of DT-MRI derived quantities computed from DWI datasets acquired with different phase encode directions will be highest in regions that are most significantly affected by EPI distortions. An effective EPI distortion correction strategy should result in a reduction of this local variability.

Experimental testing. We acquired two axial DWI datasets with phase encoding direction anterior/posterior (A/P) and right/left (R/L) respectively. In some subjects we acquired two additional datasets with opposite sign of the phase-phase encode blips resulting in either compression or expansion distortion along the same orientation. Data were acquired on a 1.5 Tesla scanner (GE Medical Systems, Milwaukee, WI) with a single-shot spin-echo EPI sequence with the following parameters: FOV = 24x24cm, slice thickness = 2.5mm, no gap, matrix = 96x96 zerofillled to 128x128, 60 axial slices. Each DTI dataset consisted of 3 images with b=0s/mm^2 and 12 images with b=1100s/mm^2 with different orientations of diffusion sensitization. An undistorted T2 weighted structural scan was acquired with a fast spin echo sequence.

All images were co-registered to correct for rigid body motion and eddy current distortion correction and all data sets were registered with a rigid body registration to a common template [7]. The EPI deformation field was computed by registering the first b=0 EPI image of each data-set to the undistorted T2 weighted structural template using a B-spline deformable registration algorithm with mutual information as the metric. This algorithm was implemented in C++ using the registration library in the Insight Segmentation and Registration Toolkit (ITK)[8]. B-spline registration was constrained to allow spatial deformation along the phase-encoding direction only. The computed deformation field was applied to all DWIs pertaining to the data-set. For each data-set non-linear tensor fitting was performed for EPI distortion corrected (C), and non-corrected (NC) DWI data. Maps of the fractional anisotropy (FA) and the trace of the diffusion tensor (TR) were calculated. The difference FA(R/L) – FA(A/P) and TR(R/L) – TR(A/P) for both the C and NC case were calculated.

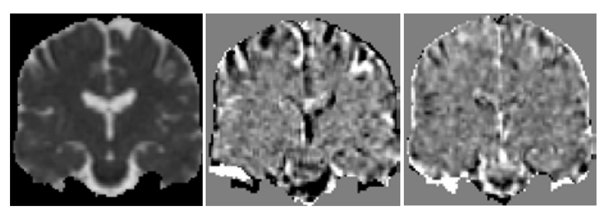
Results

EPI distortion correction reduced the variability between R/L and A/P acquisitions for both FA and TR. Representative results are displayed in Fig 1 for a sagittal view of FA. The difference FA(R/L) – FA(A/P) is shown in 1(b) for the uncorrected images and in 1(c) for the corrected images. Values range from +0.5 (white) to -0.5 (black), the grey background corresponds to 0. Note the improvement after EPI distortion correction, particularly in the genu of the corpus callosum. Fig 2 shows a coronal view of TR. The difference TR(R/L) – TR(A/P) is shown in 2(b) for the uncorrected images and in 2(c) for the corrected images. Values range from +1000 μm^2/s (white) to -1000 μm^2/s (black), the grey background corresponds to 0 μm^2/s. Note the improvement after EPI distortion correction in particular at interfaces tissue-CSF.



(a) (b) (c)

Fig. 1. FA image for anatomical reference (EPI corrected, R/L) (a); R/L – A/P difference image without EPI correction (b) and with EPI correction (c).



(a) (b) (c)

Fig. 2. TR image for anatomical reference (EPI corrected, R/L) (a); R/L–A/P difference image without EPI correction (b) and with EPI correction (c).

Conclusion

EPI distortions in diffusion weighted images (DWIs) affect the quality of brain DT-MRI. We found that a significant improvement can be achieved by registering the DWIs to an undistorted structural template with a Bspline-based image registration algorithm. This approach is relatively simple and does not require additional data, such as B_0 field maps. We propose a general framework to compare different EPI correction strategies in the context of DT-MRI. Our approach is based on the acquisition of diffusion data of the same subject with different phase encoding schemes and can be implemented without specialized hardware or software.

References

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