Correcting geometric distortions of Echo Planar Imaging using demons and reversed phase encoding

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Introduction

Echo Planar Imaging (EPI) sequences are subject to imaging artifacts, caused by subject motion, eddy currents effects and field inhomogeneity distortions (susceptibility) causing a geometric displacement of voxel intensities along the phase encode direction. Inhomogeneity correction is important to obtain an anatomical correct image which can be aligned with structural MR images. To estimate the displacement field that allows correction, we extend an existing technique, based on acquiring a full EPI sequence and one additional EPI image acquired with reversed gradient polarity along the phase encoding direction [2, 6]. The EPI with reversed gradient polarity contains the same intensity information as a corresponding image of the EPI sequence but with distortions causing voxel shifts in the reverse directions. To find the displacement field between two reversely distorted EPI's, an image registration problem is solved as in [2]. We propose to use the simpler, more efficient Thirion's demons algorithm [3, 4] and suggest a different registration pipeline for obtaining the displacement fields and name it phase reversed demons (PRD). The PRD is compared to two other correction methods that require additional MRI sequences, leading to increased scan time. These are the gradient field map (FM) [1] and the point spread function (PSF) [5]. We compare the three methods applied to five subjects. The results are compared visually and quantitatively by estimating the statistical dependence with a structural T1-weighted image. The quantitative results indicate that the (PRD) approach is competitive by being more similar with the structural image but inspection of regions in subjects also demonstrates individual cases where the other methods are favorable.

Method

Diffusion data and preprocessing: Five subjects were scanned, acquiring a structural T1-weighted (MPRAGE) (TR=1900ms and TE=2.32ms, 224 slices with 0.9mm^3 isotropic voxels) and a whole brain diffusion weighted (DWI) EPI using Twice-Refocused spin echo sequence [7] (TR=11440ms, TE=89ms, Echo Spacing=0.66ms, 61 slices, with 2.3 mm^3 isotropic voxels and GRAPPA=2), consisting of 10 b0 and 61diffusion weighted images, at b-value 1500 mm/s^2. The images were acquired on a Siemens Verio 3T MR scanner using a 32 channel head coil. We correct the DWI using a displacement field, estimated with the three approaches:
Image registration (PRD): As mentioned in the introduction, this method requires one additional b0 with same sequence parameters as the b0s of the DWI but with reversed gradient polarity. Thirion´s demons estimates the displacement field φ, as the minimizing solution to a non linear sum of squares cost function based on the differences between a displaced b0 image and a b0 image with reversed gradient polarity. To increase numerical stability of the cost function, Thirion's demons incorporate a prior in a way particularly efficient, compared to other registration implementations as argued in [3]. The displacement field for geometrically correcting EPI images is given by half the estimated displacement field, φ/2.
Avoiding local registration minima's of the cost function: A 3 stage successive registration is proposed and for each stage the solution of the previous stage is used as a start guess for the next, similarly the influence of the diffusion prior is decreased at each stage. The procedure is: 1) Estimate the field between the two background thresholded/smoothed b0 images. 2) Estimate the field between two double thresholded and smoothed b0 images using a threshold for the background and one for the Cerebral Spinal Fluid (CSF). 3) Estimate the field based on the true intensity images. Note that we only introduce intensity modulation into the image registration cost function during stage 3 of the image registration. Intensity modulation is done by multiplying the corrected EPI images with the factor (1+ (Jacobian of the field)).
The field map (FM): A double gradient echo sequence was acquired on the scanner to estimate the b0 field inhomogeneity with TR=479ms, TE1=4.92, TE2=7.38ms and isotropic voxel resolution of 3mm^3. Using the field map toolbox of SPM8 [1] the displacement field was estimated and resliced to DWI resolution.
The PSF: Point spread function was mapped using the same EPI parameters as the b0-aqcuisition with an additional sequence parameter PSF rFoV=4, displacement maps were then calculated using online software, see [5].
Method comparisons: The first b0 of the DWI sequence is corrected using the three methods and compared to the minimal distorted MPRAGE, used as the gold standard of an undistorted image. Using mutual information (MI), the MPRAGE and its affiliated brain mask shown in Figure 1 were rigidly aligned to the corrected b0 image. The mask consists of eight labeled regions allowing for region wise analysis. Within the mask, intensities were scaled to the range [0 512] and MI estimated using 512 histogram bins. MI is calculated for each subject and for each region of interest across subjects. MI is a suitable criterion that quantifies the dependence between the distributions of the two images with increased dependence suggesting higher similarity.

Results

The MI calculated within each subject revealed that the PDR method was better in 4 of 5 subjects while the FM was better in the last subject. The comparison of methods based on MI scores is plotted in Figure 1 and show that the PDR perform similarly or better than the other methods in most regions. Figure 2(a-c) allows for a visual comparison of a corrected b0 using the 3 different methods, demonstrating how similar they are. To emphasize where/how the methods actually differ, Figure 2(d) shows an absolute difference between a PSF and PDR corrected axial slice. Largest within mask differences are observed near CSF/tissue edges. Since PDR has increased MI in these regions, this could indicate increased accuracy of the PDR. To emphasize where we found PSF to be preferred, Figure 2 (e-f) compares PSF/PDR in region 5, 6 near the Petrous temporal bone. Arrows indicate where PSF had better performance.

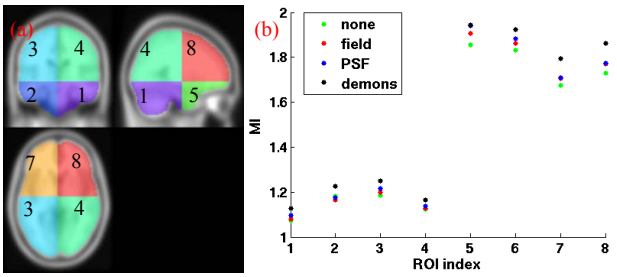

Figure 1: (a) Shows the region of interest brain mask, **(b)** The MI scores, between b0 and MPRAGE, calculated and averaged over 8 different regions.

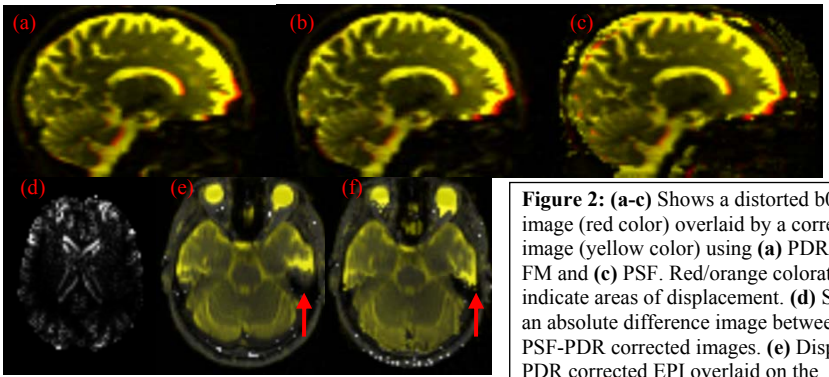

Figure 2: (a-c) Shows a distorted b0 image (red color) overlaid by a corrected image (yellow color) using **(a)** PDR, **(b)** FM and **(c)** PSF. Red/orange coloration indicate areas of displacement. **(d)** Shows an absolute difference image between PSF-PDR corrected images. **(e)** Displays PDR corrected EPI overlaid on the structural MPRAGE. **(f)** The same slice as (e) but for PSF.

Discussion

The PRD method demands less additional scan time compared to FM and PSF while achieving similar performance. Because it can be further improved, it is a viable alternative for inhomogeneity correction. We observed needs for improvements in brain region 5 and 6 where it was sometimes worse than PSF but on par with the FM (results not shown). Speaking against PSF is the sensitive to brain masking. For instance in Figure 2 (c) is shown an unmasked image with structures appearing outside the brain. About the MI measure; It is sensitive to choosing appropriate histogram bin sizes which may alter the conclusions. However, we found that conclusions were robust for both 256 and 512 number of bins. We observed magnitude difference in MI in frontal regions 5-8 compared to 1-4, possibly caused by more structural details in the back of the brains.

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