A Comparison of Techniques for Correcting Eddy-current and Motion-induced Distortions in Diffusion-weighted Echo-planar Images

Jun Haneda[1], Akifumi Hagiwara[2,3*], Masaaki Hori[2], Akihiko Wada[2],
Issei Fukunaga[2], Katsutoshi Murata[4], Koji Kamagata[2], Shohei Fujita[2,3],
Tomoko Maekawa[2,3], Ryusuke Irie[2,3], Tomohiro Takamura[2], Michimasa Suzuki[2],
Kanako Kunishima Kumamaru[2], and Shigeki Aoki[2]

The purpose of this study was to show the efficacy of dynamic field correction (DFC), a technique provided by the scanner software, in comparison to the FMRIB Software Library (FSL) post-processing "eddy" tool. DFC requires minimal additional scan time for the correction of eddy-current and motion-induced geometrical image distortions in diffusion-weighted echo-planar images. The fractional anisotropy derived from images corrected with DFC were comparable to images corrected with the "eddy" tool and significantly higher than images without correction, which demonstrates the utility of DFC.

Keywords: *diffusion-weighted magnetic resonance imaging, dynamic field correction, echo-planar imaging, eddy current*

Introduction

The use of diffusion MRI (dMRI) acquired by echo-planar imaging (EPI) has become ubiquitous in neuroimaging studies.[1] The primary advantage of dMRI is that it provides information about the microstructure and connectivity of the brain.[2–4] Despite the feasibility of dMRI by EPI, these images contain geometrical distortions and intensity variations caused by field imperfections in conjunction with a low bandwidth in the phase-encode direction. The well recognized origins of field imperfections are susceptibility-induced local gradients and eddy-current-induced large-scale perturbations.[2,5]

The field imperfection of eddy-current-induced large-scale perturbation is caused by the strong, rapidly switching diffusion encoding gradients. When the magnetic field rapidly changes, it induces eddy currents in the conducting structures of the MR scanner, which subsequently induce perturbations of the magnetic field. In most MRI applications, eddy current effects are small and insignificant. However, they can cause significant image distortion on EPI-based dMRI. This distortion causes misalignment between the acquired images because of gradients differing in strength or direction.

There is also the inevitable subject movement, especially when the scan duration is prolonged due to the acquisition of multiple *b* values and diffusion directions. These distortions may lead to serious image artifacts and misinterpretation.

A post-processing method for the simultaneous correction of eddy-current and motion-induced distortions is provided by the "eddy" tool in the FMRIB Software Library (FSL; The Analysis Group, Oxford, UK. https://fsl.fmrib.ox.ac.uk/fsl/fslwiki). The underlying algorithm assumes that the signal from two dMRI acquisitions with nearly opposed diffusion gradient directions should be identical. The main process performs outlier detection and identifies slices where the signal distortion has been caused by subject movement during the scan. These slices are replaced by non-parametric predictions that assume a Gaussian process. Yamada et al.[1] have reported that the images corrected with the eddy tool showed higher fractional anisotropy (FA) than images without correction. Because misregistration generally leads to lower FA, they concluded that the eddy tool is useful for accurate measurements on dMRI.

Alternative methods for correcting eddy-current distortions are provided by the software of commercially available MR scanners. Our scanner offers a method called "dynamic field correction" (DFC) which automatically processes the images during acquisition.

[1]Department of Radiology, Koshigaya Municipal Hospital, Saitama, Japan
[2]Department of Radiology, Juntendo University School of Medicine, 2-1-1 Hongo, Bunkyo-ku, Tokyo 113-8421, Japan
[3]Department of Radiology, Graduate School of Medicine, The University of Tokyo, Tokyo, Japan
[4]Siemens Healthcare Japan, Tokyo, Japan
*Corresponding author, Phone: +81-3-3813-3111, Fax: +81-3-3816-0958, E-mail: a-hagiwara@juntendo.ac.jp

The purpose of this study was to validate the DFC method, by comparing it to the FSL "eddy" tool.

Materials and Methods

Study subjects
This is a prospective study that was approved by our local Institutional Review Board. Informed consent was obtained from all volunteers prior to the study. Fifteen healthy volunteers (7 men and 8 women; mean age 26.1 years, age range 22–33 years) participated in this study. No subjects had a history, symptoms, or treatment for neurological disease.

MR scanner and imaging parameters
All images were obtained using a 3T MR scanner (MAGNETOM Prisma; Siemens Healthcare, Erlangen, Germany) and a 64-element phased array head coil. The dMRI protocol was based on an EPI sequence (TR/TE = 3700/69 ms, matrix size = 130 × 130, FOV = 230 × 230 mm^2, slice thickness [TH] = 1.8 mm, number of slices = 84, bandwidth = 2136 Hz/pixel, acquisition time = 4 min 53 s). Sixty-four diffusion gradient directions (b value of 1000 s/mm^2) were applied using the same parameters with phase-encoding along the antero-posterior direction. For a b value of 0 s/mm^2, phase-encoding blips of the opposite polarity were also performed.

Theory of DFC
This section provides some background information of the algorithm used for the correction of eddy-current and motion-induced distortions on dMRI.[6,7] For each acquired dMRI, eddy current-induced distortions are corrected by performing a non-affine registration to the reference image. The reference image is a non-dMRI acquisition with a given measurement protocol which typically uses a b-factor of 0. The acquired distorted dMRI is corrected by applying a four-parameter model, which includes: a) A shift due to a residual gradient in the slice-encoding direction or subject translation along the phase-encoding direction (both yielding a uniform image shift along the phase-encoding axis), b) the shear due to the residual gradient in the frequency-encoding direction (a shift along the phase-encoding axis that depends linearly on the position along the frequency-encoding direction), c) scaling due to the residual gradient along the phase-encoding direction (a shift along the phase-encoding direction that depends linearly on the position along this axis), d) higher-order deformations, using information of the system- and diffusion-direction-specific spatial patterns of the actual eddy-current fields. To cope with contrast differences between dMRI and the reference non-dMRI, the registration algorithm uses an entropy-based metric. Updating the reference non-dMRI during the image acquisition reduces the impact of subject motion on the registration procedure. Generally, dMRI deformations are expected to be non-affine on larger length-scale, except under ideal conditions.[8]

Image processing
Corrected and uncorrected images from the same scan were stored in the scanner database in digital imaging and communication in medicine (DICOM) format. After the data export, all DICOM images were converted into the Neuroimaging Informatics Technology Initiative (NIfTI) format using the "dcm2nii" tool on MRIcron (http://people.cas.sc.edu/rorden/mricron/index.html). Next, the FSL "topup" tool (v 5.0.9) was applied to images (DFC-corrected [DC]) and images (Non-corrected [NC]). The FSL "eddy" tool was then applied to NC images only for correcting eddy-current and motion-induced distortion, and the resulting images were named eddy-corrected (EC). The eddy tool estimates eddy-current induced field perturbations and head motion by a Gaussian process predictor.[1] This Gaussian process predictor offers a nonparametric approach, which retains the available data and performs an inference that is conditional on the current and local states.[9] Then, FA maps were created using the FSL "dtifit" tool from the DC, NC, and EC images. In this study, the assumption for the FA maps is that FA values are decreased by misregistration from spatial distortion in the white matter.[1] Thus, the image with the best quality should have the highest white matter FA values.

Tract-based spatial statistics
All FA maps were processed using tract-based spatial statistics (TBSS)-based voxelwise statistical analysis to quantify differences in the white matter microstructure of the DC, NC, and EC images.[10] The FMRIB58_FA standard-space image was used as the target image to non-linearly register the maps. The FA maps were next averaged to create the FA skeleton. For the exclusion of voxels from adjacent gray matter and cerebrospinal fluid, the FA value was thresholded at 0.2.[1] All aligned FA maps were projected onto the skeleton filled with the maximum FA values from a plane perpendicular to the local skeleton structure to each voxel of the skeleton.

Statistical analysis
We performed voxelwise statistics of the skeletonized FA data using nonparametric statistical thresholding (the "randomize" permutation algorithm in FSL). The function performed 5000 permutations and statistical inference using threshold-free cluster enhancement with P values < 0.05 considered to be significant after family-wise error corrected for multiple comparisons. Comparisons were performed using paired t-test between the DC and NC, and DC and EC images.[11]

Results

All images were confirmed by visual inspection to be adequate for the analysis. The NC, DC, and EC maps with a b value of 1000 s/mm^2 averaged across 64 diffusion gradients are shown in Fig. 1a–1c. The NC, DC, and EC FA maps are shown in Fig. 1d–1f. No obvious difference in quality was detected by visual assessment of isotropic and FA maps between the NC, DC, and EC images. DC images showed a significantly higher FA than NC images in all tracts by the TBSS analysis (Fig. 2). However, there was no significant difference in FA between DC and EC by TBSS.

Discussion

The present study compared a vendor-supplied method to an established FSL tool for the correction of eddy current and motion-induced image distortion in dMRI acquired by EPI. TBSS analysis showed significantly higher FA values with corrected images than those without corrections, which indicated misalignment of dMRI without correction due to spatial distortion in white matter regions. Both DFC and the FSL "eddy" tool successfully reduced these distortions.

A comparison was made first between DC and NC maps, with DC showing higher FA values, meaning that DFC effectively corrected the image distortions. Next, a comparison was made between DC and EC images, resulting in no significant differences. Collectively, these results indicate that both approaches, DFC and the "eddy" tool, yielded an equivalent level of correction. Using automated inline correction eliminates possible user errors when applying manual procedures and enables routine usage in a clinical setting (offline eddy post-processing requires longer time on a commercial computer).

A recent study showed that a reduced phase direction FOV technique using a spatially selective phase encoding gradient can offer higher DWI with improvement of the spatial resolution, and without an association with a wraparound artifact. The technique also resulted in less motion and susceptibility artifacts, which are commonly observed with a
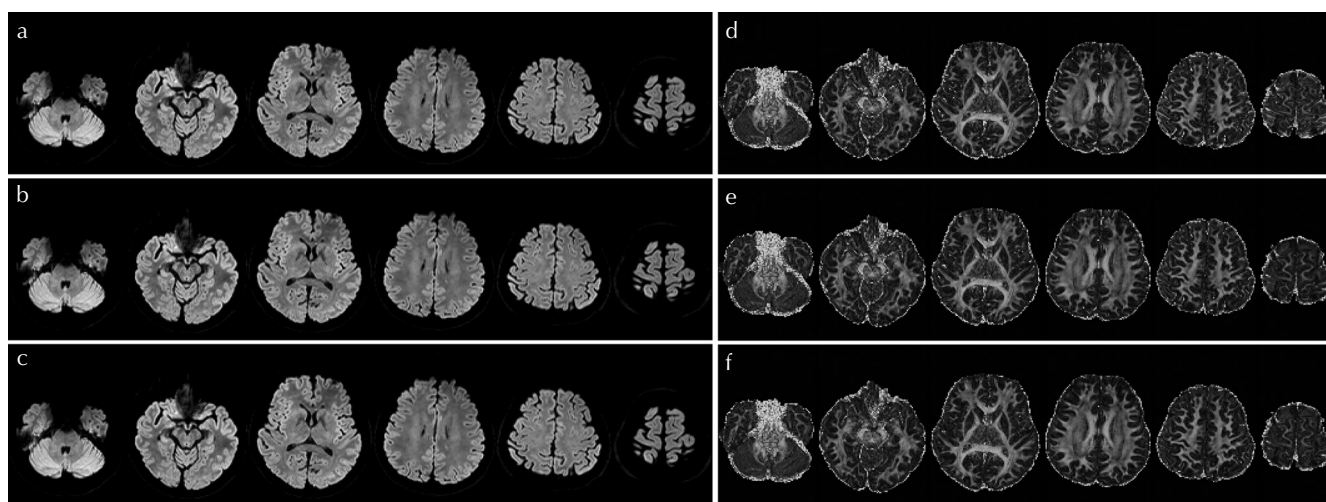

Fig. 1 Representative images from a 22-year-old male volunteer. The diffusion MRI (dMRI) with a b value of 1000 s/mm^2 averaged across 64 diffusion gradients is shown in (**a**) non-corrected (NC), (**b**) dynamic field correction-corrected (DC), and (**c**) eddy-corrected (EC). The fractional anisotropy (FA) maps are shown in (**d**) NC, (**e**) DC, and (**f**) EC.
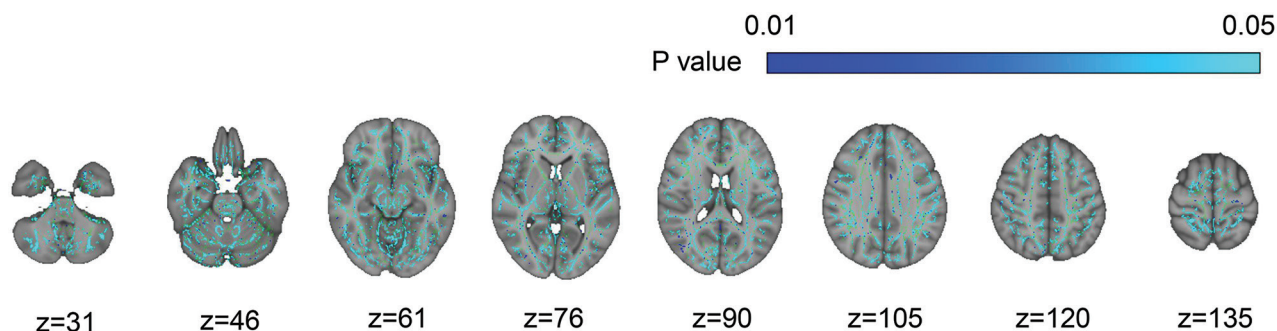

Fig. 2 Image comparison of fractional anisotropy (FA) on tract-based spatial statistics (TBSS) for different slice positions. The comparison between dynamic field correction-corrected (DC) and non-corrected (NC) images is shown. P values are shown color-coded. DC shows a significantly higher FA than NC. These data were overlaid onto the MINI152_T1_1 mm template. The significance level was set at a P value of < 0.05.

larger FOV. This technique limits the image to a smaller FOV, but may be appropriate when applied in combination with DFC.[12,13]

This study was limited to the scanning of healthy volunteers with a mean age of 26.1 years, for whom head motion may be less than that seen in older patients or children. Further investigations with different FOVs, the newer version of FSL software, and a larger number and wider age range of volunteers, including patients with suspected neurological conditions are warranted. Another limitation is that we investigated only FA as an indicator of quality on dMRI.

Conclusion

In conclusion, we have demonstrated the efficacy of DFC, which enables correction of eddy-current induced distortions comparable to the FSL "eddy" tool.

Acknowledgments

Conflicts of Interest

Katsutoshi Murata is an employee of Siemens Healthcare. The remaining authors declare that they have no conflict of interest.

References

1. Yamada H, Abe O, Shizukuishi T, et al. Efficacy of distortion correction on diffusion imaging: comparison of FSL eddy and eddy_correct using 30 and 60 directions diffusion encoding. PLoS ONE 2014; 9:e112411.
2. Jezzard P, Barnett AS, Pierpaoli C. Characterization of and correction for eddy current artifacts in echo planar diffusion imaging. Magn Reson Med 1998; 39:801–812.
3. Kamagata K, Zalesky A, Hatano T, et al. Connectome analysis with diffusion MRI in idiopathic Parkinson's disease: evaluation using multi-shell, multi-tissue, constrained spherical deconvolution. Neuroimage Clin 2018; 17:518–529.
4. Hagiwara A, Hori M, Yokoyama K, et al. Analysis of white matter damage in patients with multiple sclerosis via a novel in vivo MR method for measuring myelin, axons, and G-ratio. AJNR Am J Neuroradiol 2017; 38:1934–1940.
5. Haselgrove JC, Moore JR. Correction for distortion of echo-planar images used to calculate the apparent diffusion coefficient. Magn Reson Med 1996; 36:960–964.
6. Geppert C, Feiweier T, Janka R, Uder M, Kiefer B, Wenkel E. Reduction of image distortions in breast DWI using dynamic field correction: impact on ADC assessment. Proceedings of the Annual Meeting of the European Society of Magnetic Resonance in Medicine and Biology (ESMRMB) Leipzig, Germany, 2011; 28.
7. Feiweier T, Huwer S, Kim TH, Porter DA, Speckner T. Method and magnetic resonance system to reduce distortions in diffusion imaging. US patent 8,508,226 B2. August 13, 2013.
8. Wen Q, Basu A, Janmey PA, Yodh AG. Non-affine deformations in polymer hydrogels. Soft Matter 2012; 8: 8039–8049.
9. Hansen J, Murray-Smith R, Johansen T.A. Nonparametric Identification of linearizations and uncertainty using Gaussian process models - application to robust wheel slip control. Proceedings of the 44th IEEE Conference on Decision and Control, Seville, Spain, 2005; 5083–5088. https://ieeexplore.ieee.org/document/1582968/ (Accessed: July 13, 2018).
10. Smith SM, Jenkinson M, Johansen-Berg H, et al. Tract-based spatial statistics: voxelwise analysis of multi-subject diffusion data. Neuroimage 2006; 31:1487–1505.
11. Keihaninejad S, Ryan NS, Malone IB, et al. The importance of group-wise registration in tract based spatial statistics study of neurodegeneration: a simulation study in Alzheimer's disease. PLoS ONE 2012; 7:e45996.
12. Takeuchi M, Matsuzaki K, Harada M. Evaluating myometrial invasion in endometrial cancer: comparison of reduced field-of-view diffusion-weighted imaging and dynamic contrast-enhanced MR imaging. Magn Reson Med Sci 2018; 17:28–34.
13. Yokohama T, Iwasaki M, Oura D, Furuya S, Okuaki T. The reliability of reduced field-of-view DTI for highly accurate quantitative assessment of cervical spinal cord tracts. Magn Reson Med Sci 2018; 18:36–43.