Effects of Preprocessing on Goodness of Fit Measures in Premature Neonatal DTI

D. Morris[1], R. Nossin-Manor[1], M. J. Taylor[1], and J. G. Sled[2,3]

[1]Diagnostic Imaging, The Hospital for Sick Children, Toronto, Ontario, Canada, [2]Physiology Experimental Medicine, The Hospital for Sick Children, Toronto, Ontario, Canada, [3]Department of Medical Biophysics, University of Toronto, Toronto, Ontario, Canada

Introduction: Diffusion tensor imaging (DTI) is a non-invasive tool for investigating white matter development in the brain. As an EPI-based technique, it has specific artifacts associated with subject motion, susceptibility, field inhomogeneity and gradient imperfections. Various processing methods have been proposed to correct these artifacts [1-3]. However, correction is particularly challenging when imaging premature neonates without sedation as this population is particularly prone to motion. In this abstract, we define a DTI pipeline for premature neonates to mitigate artifacts due to subject head motion, brain pulsation, distortions from eddy currents, and B_0 field inhomogeneity. We then investigate the effect of each processing step on the goodness of fit of the resulting tensor model.

Subjects: DTI was successfully performed on 19 premature neonates (mean post conceptional age at birth 28.2±1.7, at scan 29.6±1.8 weeks) as part of a longitudinal multi-modal study. Subjects were recruited from the neonatal intensive care unit with written, informed parental consent and in compliance with hospital ethics. Infants were swaddled and fed prior to imaging; no sedation was administered.

Imaging: Images were acquired on a GE 1.5T system (Signa Excite HD 12M4) using a third party MR compatible incubator and head coil (Advanced Imaging Research, Cleveland, USA). Twice refocused spin echo planar DTI [4] was acquired using: 2D axial oblique slices, FOV=205mm, with 1.6 mm cubic voxels. TR/TE/FA = 15s/85ms/90°. 3 non-diffusion and 15 non-collinear diffusion-weighted volumes with b=700 s/mm^2 and were acquired. A field map was also acquired using two spoiled gradient echo (SPGR) images at TE = 8 and 12.7 ms, TR/FA 24ms/19° and 2.5x2.5x3 mm^3 resolution. Both sequences had full brain coverage.

Processing: Pipeline A (AR-B0-OR): Diffusion weighted volumes were aligned to a non-diffusion weighted volume using affine registration (AR), to correct for subject motion and first order geometric distortions caused by eddy currents. A B_0 field map was calculated from the phase difference between the two SPGR images, followed by application of a phase unwrapping algorithm (FSL *PRELUDE* [5]). The field map was unwrapped then used to correct B_0 field inhomogeneity and susceptibility induced distortions (B0) using FSL *FUGUE* [5]. At each processing stage, diffusion tensors were estimated using a standard non-linear fitting algorithm as well as a robust fitting procedure (*RESTORE*) that incorporated voxelwise outlier rejection (OR) [2,6]. For each tensor fit, the mean square residual was calculated as $\Sigma_{i=1}^{15}[S_{D,i}/S_0 - \exp(-b\cdot\vec{g}_i\hat{D}\vec{g}_i^T)]^2/(15-n_O)$ where $\vec{g}_i$ is the diffusion gradient vector, $\hat{D}$ is the tensor estimate, $S_{D,i}$ and S_0 are the diffusion and non-diffusion weighted signals, and n_O is number of outlier points excluded from the model. Fractional anisotropy (FA) maps were also calculated. Pipeline B (OF-AR-B0): *RESTORE* was used to identify outlier voxels in the uncorrected data. These were then replaced by a weighted local average using a 2 mm full-width half-maximum filter. This "outlier free" (OF) dataset was then put through affine registration and B_0 field map correction before nonlinear tensor fitting. Using an FA map, regions of interest (ROIs) were drawn over the genu and splenium of the corpus callosum (GCC, SCC), the internal capsules (IC) and in low anisotropy frontal white matter (FWM). Within each ROI, the mean FA was calculated at each processing stage. To investigate whether processing steps had differential effects on high and low anisotropy regions, data were also divided into voxels with FA > 0.22 and FA < 0.22.

LEGEND:

AR – affine registration	**B0** – B_0 distortion correction
OR – outlier rejection	**OF** – outlier free

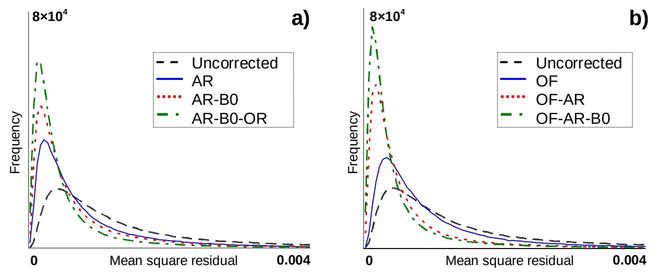

Fig 1. *Histograms of mean squared residuals of pipeline A (a) and B (b). Both pipelines show continuous improvement with each processing step.*

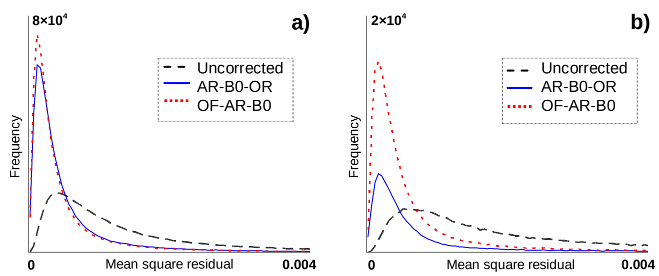

Fig 2. *Histograms of comparing mean squared residual in both pipelines for all voxels (a) and high FA voxels (b). Outlier correction as a first step improves tensor fit especially for high FA voxels.*

Results: Figure 1 shows mean square residual histograms for both the AR-B0-OR (Fig 1a) and OF-AR-B0 (Fig 1b) pipelines across all subjects. Both show a continual decrease in residuals at each processing step. Similar patterns were seen at the single subject level. Figure 2 shows a comparison between the two pipelines when a) all voxels are included and b) when only voxels with FA > 0.22 are included. Results for low FA voxels were similar to Fig 2a. Figure 3 shows the mean FA in the ROIs before correction and after each processing step.

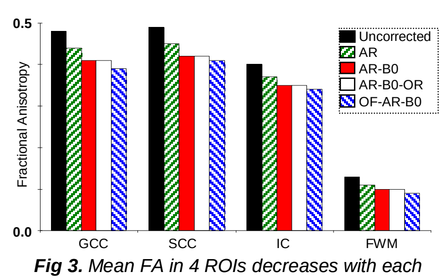

Fig 3. *Mean FA in 4 ROIs decreases with each processing step.*

Discussion and Conclusions: The described pipelines effectively improve the integrity of DTI data in preterm neonates with processing step. Noise and artifacts in preterm DTI data tend to inflate estimated FA values, especially in more anisotropic regions. Performing outlier replacement prior to other corrections reduces residuals further in high FA regions. This immediate removal of outliers prevents their spread across several voxels during interpolations that occur in subsequent steps. This technique can be extended to any subject population prone to motion.

References:

1 Mangin *et al*. Med Imag Anal 6:191, 2002
2 Chang *et al*. MRM, 53:1088, 2005
3 Jezzard *et al*. MRM 34:65, 1995
4 Reese *et al.* MRM, 49:177, 2003
5 Smith *et al*. Neuroimage 23(S1):208, 2004
6 Cook *et al*. ISMRM 14:2759, 2006