# Testing the reproducibility of Diffusion Tensor Imaging at 3.0T

**J. Frandsen[1], L. Østergaard[1], and E. B. Vedel Jensen[2]**

[1]Center for Functionally Integrative Neuroscience, Aarhus University Hospital, Aarhus C, Aarhus, Denmark, [2]The T. N. Thiele Centre of Applied Mathematics in Natural Science, Aarhus University, Aarhus C, Aarhus, Denmark

**Purpose**

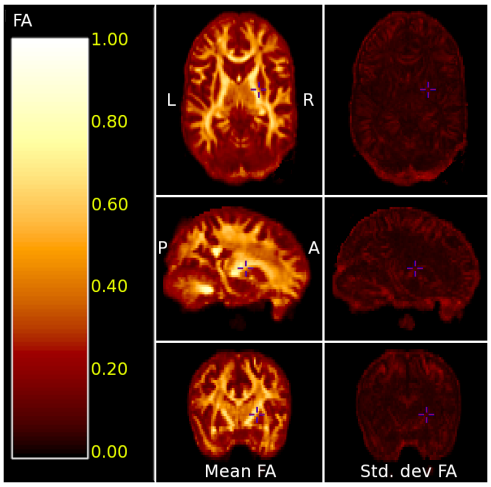
Diffusion-weighted imaging (DWI) may be utilized to non-invasively probe brain microstructure. Utilizing the restriction of water diffusion perpendicular to white matter axons, diffusion anisotropy may be measured and characterized by a 3x3 matrix, the diffusion tensor [1]. From this tensor several scalar indices are derived and this includes typically the apparent diffusion coefficient (ADC) and an index of anisotropy such as the fractional anisotropy (FA) [2]. FA and ADC are among the scalar indices used for comparing groups of healthy controls with groups of neurodegenerative deceases such as multiple sclerosis [3] and schizophrenia [4]. Furthermore, axonal fiber tracking may be applied to the full tensor, reconstructing the course of major white matter fiber bundles [5]. In this study we assessed the maximum attainable reproducibility of scalar indices of water diffusion and of white matter fiber tracking by comparing repeated, co-registered measurements in the same subject.

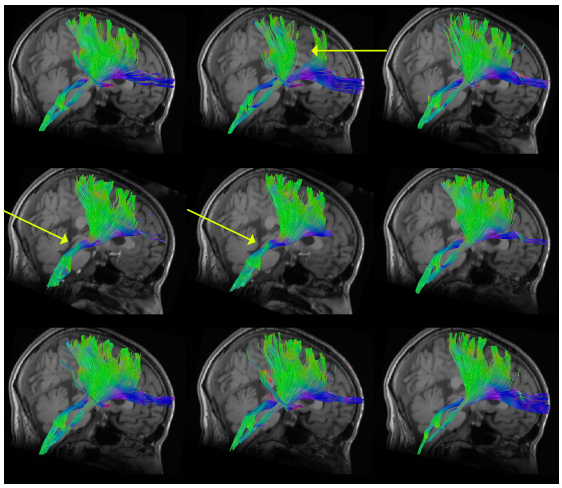
**Methods**

One healthy volunteer (38 year old male) with no history of neurological or psychiatric symptoms was scanned 20 times within 21 days. $T_1$ and DWI sequences were obtained with a 3.0 T GE Signa System MR scanner (GE medical systems, Milwaukee, USA) using a birdcage head coil. Diffusion Tensor Imaging (DTI) was performed with a double spin echo single shot EPI sequence. The diffusion encoding scheme consisted of 26 directions isotropically distributed in space, b =1000 s/mm2. In addition 6 b-factor=0 s/mm2 image were acquired. No gating was applied. The maximum gradient strength was 36 mT/m. 51 slice locations of slice thickness 3.0 mm were acquired, using 24 cm FOV in a 128 by 128 image matrix. TR/TE=12500/71 ms with two repetitions, resulting in a acquisition time of 14 minutes. $T_1$ weighted imaging parameters were TI/TR/TE=750/6.2/2.8 ms, 256 by 256 image matrix, 146 contiguous slices with a thickness of 1.2 mm, acquisition time 11 minutes. Total scanning time was hence 30 minutes. For each complete scanning the volunteer was taken out of the scanner and repositioned in the scanner. The diffusion weighted images were corrected for any signal intensity drift, diffusion tensors were calculated using a linear least squares regression method and the indices ADC and FA were calculated. Each dataset was co-registered to a common template. Average and standard derivation FA maps were furthermore calculated.

**Results**

The FA maps below left show the mean of the co-registered FA images (n=20) and the corresponding standard deviations. These were typically of the order of 5%. The largest deviations were found at gray/white matter and gray matter/CSF interfaces. We ascribe these to small misalignments due to head movements during the scan and to partial volume effects. We also speculate that registration may be sub-optimal due to the inherent image artifacts of the DTI sequence: These artifacts depend on the diffusion gradient gradient relative to the orientation of the head, limiting the precision of alignment.



Tractography was applied on all co-registered DTI data sets with identical regions of interest (ROI). The image on the right shows the result of tracking the corticospinal fibers superimposed on the $T_1$ weighted images on a subset of 9 arbitrary selected dataset. The color coding of the fiber tracts represents the local orientation. Note that even with moderate to long scan times and a controlled setup, the fibertracking results may differ among measurements within the same subject (yellow arrows).

**Conclusion**

Preliminary results indicate that FA data has an intrinsic level of noise of roughly 5% due to realignment, biological and scanner noise, using the parameters described above. Areas near tissue interfaces are particularly prone to errors, even with accurate alignment. These effects seem to affect reproducibility of fiber tracking results. We speculate that cardiac and respiratory gating may improve this reproducibility.

**References**

[**1**] Basser et al, 1994, Estimation of the effective self-diffusion tensor from the NMR spin echo, J. Magn. Reson. B 103: 247-254. [**2**] Basser and Pierpaoli, 1995, Elucidating tissue structure by diffusion tensor MRI. In Book of Abstracts, Third Annual Meeting of the Int. Soc. Of MRM, Vol 2. p900 [**3**] Thivard et al, 2007, Diffusion tensor imaging and voxel based morphometry study in amyotrophic lateral sclerosis: relationships with motor disability, Journal of Neurology, Neurosurgery, and Psychiatry;78:889-892. [**4**] Kubichki M et al, 2005, DTI and MTR abnormalities in schizophrenia: Analysis of white matter integrity, NeuroImage 26 1109-1118. [**5**] Mori et al, 1999, Three-dimensional tracking of axonal projections in the brain by magnetic resonance imaging, Annals of Neurology 45 265-269.