Dual Tensor Tracking in Low Angular Resolution Diffusion Weighted MRI

M. W. Caan[1,2], M. M. van der Graaff[1], S. D. Olabarriaga[1], C. A. Grimbergen[1], L. J. van Vliet[2], and F. M. Vos[1,2]
[1]Radiology, Academic Medical Center, Amsterdam, North-Holland, Netherlands, [2]Imaging Science and Technology, Delft University of Technology, Delft, Netherlands

INTRODUCTION. Single tensor fiber tracking is biased in regions where fibers are crossing. A dual tensor model may facilitate tracking through such complex tissue regions. A higher diffusion weighting b>>1000 s.mm^{-2} and many gradient directions are then needed (HARDI, High Angular Resolution Diffusion Imaging), requiring long scanning times. In comparitive studies to brain diseases, limited scanning time may be available, such that higher order diffusion models are out of range.

We propose to estimate a dual tensor model on an entire cohort with low diffusion weighting and a limited number of gradient directions. Diffusion attenuation profiles of multiple subjects are regarded as realizations of a single underlying fiber distribution. Non-rigid coregistration ensures spatial correspondence. Increased angular resolution is ensured by random subject positioning in the scanner, as well as by anatomical heterogeneity.

METHOD. Diffusion Weighted Images (DWIs) of all subjects are to be coregistered to a common reference space. We follow an population based atlas building method, that coregisters all pairs of datasets using a non-rigid viscous fluid model by optimizing the mutual information criterion computed on the single tensor components [1].

DW signals need to be correctly treated during transformation to pertain corresponding inter-voxel orientation. We choose to adapt the diffusion weighting vectors $\mathbf{q_j}$ based on the deformation,

$$\mathbf{q_j} = \sqrt{b} \cdot \mathbf{R}_{\text{nonrigid,j}}(x,y,z) \cdot \mathbf{R}_{\text{affine,j}} \cdot \mathbf{g_j} \qquad (1)$$

with b the diffusion weighting and gradient directions $\mathbf{g_j}$. The rotational component of the affine transformation $\mathbf{R}_{\text{affine,j}}$ is applied globally, while the rotation due to non-rigid transformation $\mathbf{R}_{\text{nonrigid,j}}(x,y,z)$ is applied per voxel (x,y,z). The advantage of this reorientation method is that both $\mathbf{R}_{\text{affine,j}}$ and $\mathbf{R}_{\text{nonrigid,j}}(x,y,z)$ will be dispersed among subjects. Random patient positioning in the scanner and inter-subject variations in bundle trajectories contribute to this dispersion. The aggregated set of gradient directions over all subjects will be more densely sampled. The combined set of multiple subjects is thus sampled at a higher angular resolution than the original acquisition protocol.

A dual tensor model with an isotropic compartment models the diffusion weighted signal $S_j(\mathbf{q_j})$:

$$\frac{S_j(\mathbf{q_j})}{S_{j,0}} = f_1 \exp(-\mathbf{q_j}^T \mathbf{D_1} \mathbf{q_j}) + (1 - f_1 - f_{iso}) \exp(-\mathbf{q_j}^T \mathbf{D_2} \mathbf{q_j}) + f_{iso} \exp(-\mathbf{q_j}^T \mathbf{q_j} D_{iso}) \qquad (2)$$

$S_{j,0}$ is the signal without diffusion weighting for subject j, $f_{...}$ are volume fractions and D_{iso} is the isotropic diffusion. The tensors $\mathbf{D}_{1,2}=\mathbf{R}_{1,2}\mathbf{E}\mathbf{R}_{1,2}$ with eigenvalue matrix $E = \text{diag}(\lambda_{\parallel}\ \lambda_{\perp}\ \lambda_{\perp})$. Rotation matrices $\mathbf{R}_{1,2}$ are parameterized by Euler angles. Levenberg-Marquardt optimization of the least-squares difference between signal and model is performed. WM fibers are reconstructed using a single and dual-tensor streamline tracking method, the latter allowing for branches [2].

RESULTS. 11 healthy subjects were scanned twice with a time interval of six months on a 3T Philips scanner, such that 22 scans were available. The diffusion weighting was b=1000smm^{-2}, 32 gradients directions were used, the spatial resolution was 2.0x2.0x2.2 mm. One S_0-image was acquired. All 22.22=484 pairs of scans were coregistered and average deformation fields were computed. These cpu and storage demanding computations were performed within a virtual laboratory on the Dutch Life Science grid. [3]

All DWIs were warped to atlas space and rotated diffusion weighting vectors were stored per voxel. A single and dual tensor model was fit in each intercranial voxel based on 484 DWIs.

Within the mesencephalon, fibers are known to cross through the decussation of the superior cerebellar peduncle (dscp). In the single tensor atlas, the red dscp voxels suggest a left-right orientation (Fig. 1a). In the dual tensor atlas, two distict orientations are observed (Fig. 1b). Superior cerebellar peduncle (scp) fibers were then tracked. While single tensor atlas fibers proceeded ipsilaterally (Fig. 1c), dual tensor fibers crossed through the dscp and traversed contralaterally (Fig. 1d). In single tensor tracking only the two caudal ROIs were included, to prevent a void tracking result.

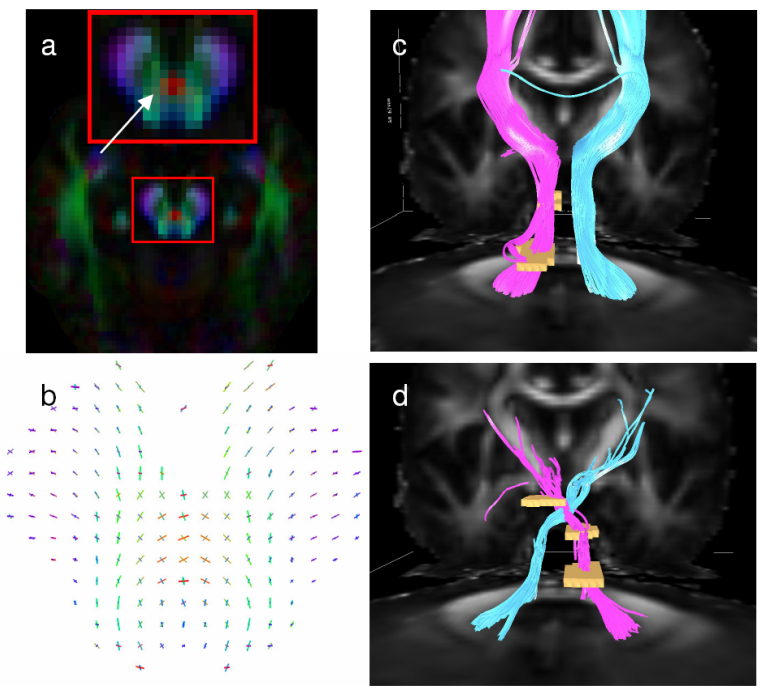

Figure 1. (a) Axial slice of the single tensor atlas, denoting FA and orientation, with mesencephalon (inset) in which the decussation of the superior cerebellar peduncle (dscp) is seen in red (arrow). (b) Principal dual tensor atlas orientations. Note the crossing fibers in the dscp. (c) Single and (d) dual tensor tractography.

CONCLUSION. We succesfully tracked crossing fibers in a dual tensor atlas, estimated from data with low diffusion weighting (b=1000smm^{-2}).

ACKNOWLEDGEMENTS.
This work was sponsored by the BiG Grid project with financial support from the NWO.

REFERENCES.

[1] Van Hecke et al., NeuroImage 43: 69-80,2008.
[2] Toussaint et al, Proc ISMRM 1339, 2005.
[3] Olabarriaga, Healthgrid, 2008.