Computing Strings of Average HARDI Models Using Procrustes-Based Fibre Alignment

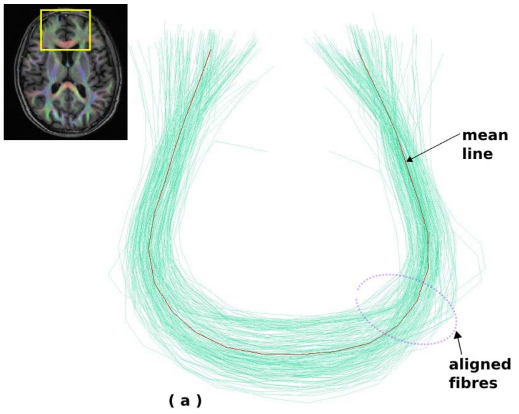
I. Kezele[1], C. Poupon[1], M. Perrin[2], Y. Cointepas[1], V. El Kouby[1], F. Poupon[1], and J-F. Mangin[1]
[1]CEA NeuroSpin, Gif sur Yvette, France, France, [2]GE Healthcare, Buc, France

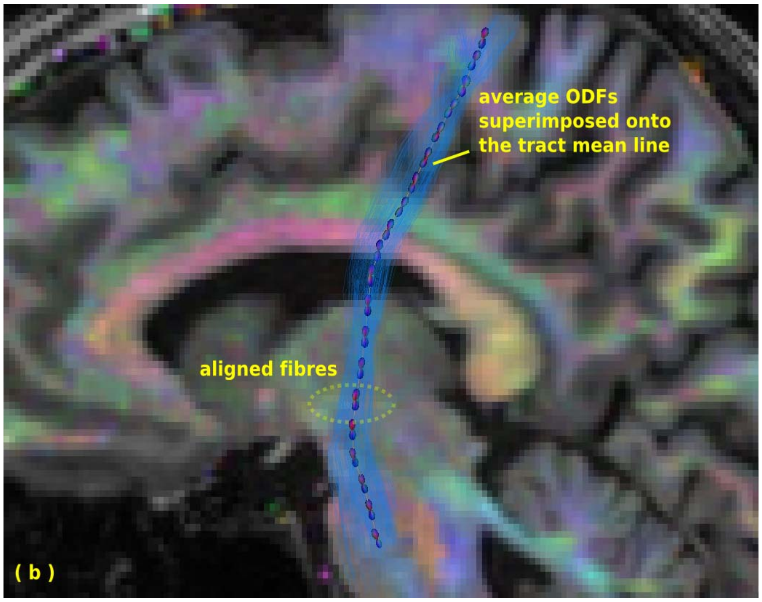
Introduction. In order to assess brain white matter (WM) changes on a microscopic scale (e.g., concerning the cyto- or myelo-architectonics) [1], morphometry techniques based on diffusion spectrum imaging or its derived methods (e.g., high-angular resolution diffusion imaging, HARDI) should be addressed. Further, to circumvent the ill-posed problem of inter-image alignment, we advocate the idea of directing the concept of object-based morphometry [2] to WM entities, such as fibre tracts. To achieve adequate representation of WM fibre tracts, the tract assigned attributes should be thought of in such a manner that they remain objective and that they do not reflect, or at worst, reflect only weakly the effects of deficient image resolution, like partial volume effects. To satisfy these demands, we put forward the idea of accentuating the information along tract mean lines, similar to [3]. Further, to avoid discarding potentially useful information away from the tract mean line while conforming to the abovementioned requirements, we introduce HARDI weighting inside the planes perpendicular to the mean line, and exploit all the information available. The attributes we opt for are the orientation distribution functions (ODFs) along mean lines, obtained from the weighted averages of HARDI data. In this text, we describe a method for mean line extraction and tract attribute definition. This method is conceived with the idea of performing robust tract-based morphometry.

Background. Recent diffusion-related studies of local WM morphometry utilised WM descriptors that were mostly based on diffusion tensor imaging (DTI) and the associated index of fractional anisotropy (FA) [3,5]. These studies aimed at global investigation of localised changes of WM. For example, the authors of [3] performed a robust voxel-based analysis that took advantage of the effective FA independence on local fibre orientation and insensitivity of tract skeleton-based approach to imperfect inter-image alignment. Likewise, our method makes use of diffusion analysis along the tract skeleton. However, the mean line that we define derives directly from the tract fibre constellation, thus exploiting the knowledge of the exact geometry of the tract's interior. This way it perhaps represents a more natural notion of the tract mean line, compared to the FA-based skeleton. Further, the tract assigned attributes that we introduce result from HARDI, and thus bear more affluent local information. The definition of the tract mean line relies on the Procrustes analysis [5] of tract fibres, and exhibits some resemblance to the work described in [6]. Nonetheless, we improve the approach of [6] by implicitly allowing for some non-rigid transformation, letting the fibre-based pseudo-landmarks slide along a pre-defined set of directions [5]. The tract based morphometry is to be performed in a dedicated coordinate system along the mean line. The partial volume problem near the tract edges is efficiently reduced by weighting the HARDI data by tracked fibre [7] density. The use of mean-line also resolves the problem of correspondence of local fibre orientations across subjects, since we translate the analysis from a global brain-based referential to a tract-dedicated referential. The problem of tract start and end points [6] remains partially unresolved, however we believe that if we define tract sub-regions that contain multiple points/attributes along the tract-dedicated coordinate system, defining the start and end planes by neighbouring grey matter (GM) mathematical or anatomical landmarks will be sufficiently accurate.

Methods. *Image Acquisition and Processing*. Twice refocused DW-SS-SE-EPI (TE/TR=100.2ms/19s, BW=200KHz, FOV=24cm on a 128x128 matrix, TH=2mm, 60 axial slices) was employed to obtain T2-weighted (b=0) and HARDI (b=3000s/mm², 200 directions) data of a healthy volunteer, on a GE Signa 1.5T system (GE Healthcare, Milwaukee). Rapid 3D echo gradient with inversion recovery (TE/TR/TI=2/9.9/600ms, BW=12,5KHz, FOV=24cm on a 256x192 matrix, TH=1.2mm, 128 axial slices) was used to acquire anatomical T1-weighted data. Diffusion-weighted data were corrected for geometric distortion related to gradient nonlinearity and susceptibility effects. After the ODF at every voxel had been estimated through Funk-Radon transform [8], a probabilistic fibre tracking algorithm [7] was performed inside a WM brain mask (obtained from the histogram analysis of the T1-weighted image), to track all the fibres starting at two regions of interest (ROIs), defined on the entire left and right cortical GM. *Fiber Tract Definition*. Fibres were clustered into 150 major fascicles using an algorithm based on WM connectivity patterns [9]. Fibres in each tract were filtered by length and distance to all other fibres to discard 10% of extreme values (of length and distance, each). For a particular tract, fibre density at each voxel was defined as the number of fibre points passing through that voxel. To extend the scale of assigned densities, fibres were supersampled prior to density calculation. *Mean Line and Mean ODFs*. For each tract, fibre pseudo-landmarks were defined by sampling each fibre at equidistant points whose number was adapted to the approximate tract curvature and the mean line was found through 3D Procrustes-based analysis, minimizing the "squared Laplacian" energy and allowing for landmark sliding along the tangent direction at each landmark [5]. Mean ODFs along the mean line were found as follows: first, thin cylindrical ROIs perpendicular to the mean line tangents at each landmark and of a radius superior to the maximal tract radius were defined; then, HARDI associated to any tract point contained inside each cylinder (where fibre density of the tract was nonzero) were rotated corresponding to the Procrustes derived rotation of a particular fibre from which the point originated (with HARDI being geodesically interpolated); rotated HARDIs were weighted by fibre density and averaged; finally, ODFs at each mean line point were estimated upon thus averaged HARDI data using fast Funk-Radon transform [10], by decomposing the signal onto a spherical harmonics basis (max order=4).

Results. The method and results are illustrated in Figure 1. Depicted are: (a) a callosal tract with fibres aligned as described in Methods above, together with the estimated tract mean line, and (b) a part of cortico-spinal tract, with aligned fibres, tract mean line, and the estimated average ODFs along the mean line. The referential shown in (a) and (b) is obtained my merging the anatomical T1-weighted image with the DTI derived image of principal directions of diffusion (red/green/blue=x/y/z).





Bibliography: 1. Ashburner et al., HBM, 6:348-57(1998); 2. Mangin et al., TMI IEEE, 23(8):968-82(2004); 3. Smith et al., NIMG 31:1487-1505(2006); 4. Kochunov et al., NIMG 35 :478-87(2007) ; 5. F.L. Bookstein, MMBIA Proceedings, 279-89(1996); 6. Corouge et al., MedIA 10 :786-798(2006) ; 7. Perrin, IPMI Proceedings (2005); 8.Tuch MRM 52:1358-72(2004), 9. El Kouby at al., MICCAI Proceedings, 196-204(2005); 10. Descoteaux et al., MRM 58(3):497-510(2007).