Separation of White Matter Fascicles From Diffusion MRI Using Φ-Functional Regularization

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Introduction

White matter mapping using Diffusion Tensor Imaging (DTI) has shown to give erroneous results in inhomogeneous voxels containing crossing white matter fibers (1). In such voxels DTI usually results with one calculated orientation, most probably different than any of the real fiber orientations. This error affects most of the fiber tracing methods by diverting the tract when it passes through inhomogeneous voxels. We propose to use a multiple diffusion tensor approach (2) combined with Φ-functional regularization to provide multiple orientations per voxel. We test this algorithm both on synthetic data and experimental data.

Theoretical Background

Eq. [1] defines the net signal decay for multiple neuronal fiber orientations as the weighted sum of diffusion tensors (2), where the coefficients *f* are the relative weights of each tensor and *q* includes the diffusion gradient vector. The diffusion tensor, D, is further described in Eq. [2] as its spectral decomposition to eigenvectors (*U*) and eigenvalues (λ). Eq. [3] defines the Φ-functional (3), which regularize the field of multiple tensors (Ω). The first term of the functional is the fitting term. This term aims to equate the measured signal decay to the modeled signal decay. The second term of the functional is the regularization term. We assume that for a single fiber, the change in *U*, λ and *f* over neighboring voxels is smooth. The minimization of this cost functional *L* via the Euler-Lagrange Partial Differential Equations (PDEs) and the gradient descent procedure, leads to an anisotropic diffusion-like flow, which reduce the orientation variation between neighboring voxels in homogeneous regions while keeping significant sharp jumps in orientation (i.e. preserving discontinuities) in transition and/or crossing areas. Regularizing the eigenvectors requires special gradient calculation techniques in order to take into account the symmetric nature of the decay signal (4). The minimization that is realized via this gradient descent flows preserves the ellipsoid shape of each tensor, and the anisotropic nature of each ellipsoid. This procedure results with multiple orientations describing each voxel.

[1] $$E(q_k) = \sum_i f_i \exp(-q_k{}^T D_i q_k \tau)$$

[2] $$D_i = \sum_j \lambda_j^i U_j^i (U_j^i)^T$$

[3] $$L(f,\lambda,U) = \int_\Omega \left(\alpha_1 \sum_k (Signal(q_k) - E(q_k))^2 + \alpha_2 \Phi_1(|\nabla f|) + \alpha_3 \Phi_2(|\nabla U|) + \alpha_4 \Phi_3(|\nabla \lambda|)\right) d\Omega$$

Methods

Synthetic data consisted of an image slice in which two fibers aligned 90^0 to each other crossed in some voxels (see Figure 1). The data included a set of 99 synthetic diffusion weighted images (DWI) corresponding to a 99 gradient direction scheme distributed equally over a sphere. The signal in each synthetic DWI was produced using Eq. [1] with the following parameters: b value of 1,000 s/mm^2 and the three diffusion eigen values were: 1.5×10^{-3}, 0.4×10^{-3} and 0.4×10^{-3} mm^2/s for each fiber.

Experimental Data: Excised spinal cords were scanned on a 7T spectrometer. Two sections of a freshly excised cervical pig spinal cord were placed crossing at about 45 degrees. Diffusion experiments were performed using a PGSE sequence with the following parameters: TR/TE= 2000/200ms, Δ/δ=150/40ms. The field of view (FOV) was 5cm, matrix size was 32 x 32 and slice thickness was 15mm. Gradient strength was 0.14 G/mm resulting in b value of 1,725 s/mm^2 measured in 31 non-collinear gradient directions.

Results and Discussion

Figure 1 shows conventional DTI analysis and the regularization algorithm results for the synthetic data. The diffusion tensor analysis, providing single fiber orientation per voxel, failed to reproduce the original fiber direction in areas of crossing fibers. By contrast, in the same voxels, the proposed regularization algorithm reproduced the exact orientation of the two fibers. Figure 2 shows the regularization algorithm results for the experimental data. In the crossing region of the two spinal cords segments the algorithms nicely reproduced the directions of the fibers in most pixels. The results suggest that the regularization algorithm enables the separation of two crossing fibers within the same voxel. Analysis of diffusion tensor data is usually done by minimizing the difference between the signal decay and a model containing a single or multiple diffusion tensors (1,2). In inhomogeneous voxels, such fitting without regulating might stumble over many local minima, which in turn may cause erroneous results. The regularization constrains the fitted tensor(s) within a voxel to be smooth with neighboring voxels thus reducing the chance of falling into local minima in the fitting procedure. This can be viewed as a noise reduction procedure in areas of voxels containing single fiber orientation. This becomes more important in voxels containing multiple fiber orientations, where the chance of falling into local minima is high. The combined contribution of the fitting term and the regularization terms is allowing change of orientations inside an inhomogeneous area (voxels with multiple fiber orientations) and yet to connect smoothly to homogenous areas (pixels with single fiber orientation). Both the synthetic and experimental data show accurate results for voxels containing only one fiber. In such voxels the orientation extracted is very similar to DTI, yet much smoother. Achieving smoothness of eigenvalues, eigenvectors and relative weights results in smoothness of the fiber delineation.

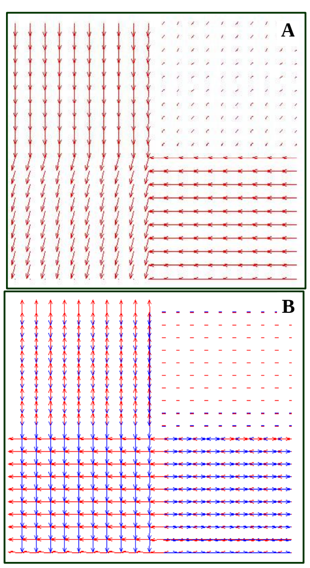

Figure1: synthetic data: DTI result (A). Regularization result, gives 2 orientations for each voxel (B)

Conclusions

We offer an approach for fitting white matter microstructual parameters (as orientation and the diffusion tensor eigenvalues) even in voxels containing inhomogeneous white matter bundles. Combining this approach with existing DTI based fiber tracing methods may help in delineating fiber tracts even in voxels containing multiple fiber orientations by choosing the fiber orientation that will be most suitable for the given tract reaching this voxel.

References

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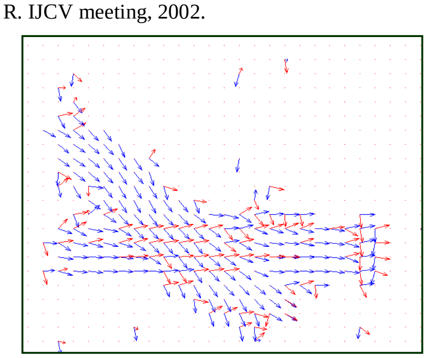

Figure 2: Real data: For each voxel. Blue arrow represents the tensor with largest partial volume (f>0.5). Red arrow represents tensor with second substancial partial volume (f>0.3).