Non-Rigid Registration of Diffusion Weighted Images using Fibre Orientation Distributions

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Introduction

Diffusion weighted imaging (DWI) provides a non-invasive method to analyse, *in vivo*, the microscopic structure of oriented tissue such as white matter. Using high angular resolution diffusion imaging (HARDI) [1], techniques such as Constrained Spherical Deconvolution [2] (CSD) and Q-Ball [3] are able to extract fibre orientation distributions (FOD). FOD based tractography techniques can then be employed to investigate white matter connectivity. However, in order to compare fibre orientations and connectivity between individuals, spatial normalisation is required. High resolution T1 weighted (T1W) images are often used to estimate inter-subject correspondence. However, very little structural information is present in the homogenous intensity of T1W white matter and therefore registration of deep white matter structures is unguided. Using T1W images for normalising DWI data requires non-rigid coregistration between DWI and T1W to correct geometric distortions often present in DWI. Other registration techniques have focused on registering either scalar information computed from the diffusion tensor (DT) [4], or the DT itself [5]. The estimation of DT becomes less reliable at high b-values required for techniques such CSD, and therefore using the DT for spatially normalising HARDI data is less than ideal. In addition, the DT does not adequately model regions with crossing fibres. In this study we propose a registration method using a novel similarity measure that optimises directly on FODs. Unlike existing techniques, this exploits the additional information provided in voxels with crossing fibres, and is independent of the model used (e.g. Q-Ball or CSD).

Method

Registration was performed using a B-spline based free form deformation (FFD) technique implemented in the Insight Segmentation and Registration Toolkit (ITK) framework. FODs were transformed using a method similar to [6]. First, the FOD was sampled with a number of equally distributed directions to produce a set of Cartesian vectors. Second, the Jacobian matrix J for each voxel was computed from the local deformation field. Third, the sampling vectors were transformed by applying J. Fourth, the FOD was estimated from the set of transformed vectors.

The deformation field was initialised with a bulk affine transform computed from an initial rigid and affine registration. Transformation of the FOD was conducted at every iteration for rigid and affine registrations, however during non-rigid registration, a different transform is applied at every voxel and therefore to reduce computation time, each FOD was transformed only once at the start of the iteration (which involves the optimisation of each control point). This transform was based on the resulting deformation of the previous iteration. Optimisation was performed by regular step gradient descent and sought to minimise the sum of the least squares difference of the FOD spherical harmonic coefficients of all voxels. To reduce the number of sampling vectors required to sufficiently characterise the FOD, and eliminate unwanted noisy peaks, registration was performed on FOD data with maximum harmonic order (lmax) of 4. The combined affine and deformable transformation was then applied to FOD data with a larger lmax of 8.

Experimental Methods and Results

An experiment to assess the algorithm's ability to reach the global minimum solution was performed by registering an individual image to itself starting from a perturbed deformation field. The initial deformation was the computed result of a registration to a different subject and therefore represented a realistic transformation. FODs were computed using MRtrix [7] from HARDI data acquired on a Siemens 3T scanner, with b=3000s/mm2, 60 DW directions and a 2.3mm isotropic voxel size. The self-registration was performed on eight subjects, using control point (CP) spacing of 10 mm and 30 iterations. Average errors were computed for those control points that lay within a tissue mask for both the initial deformation and the resulting self-registration. The results were 1.29±2.35 mm and 0.33mm±0.47 respectively. The errors for the top 1% of the most severely deformed control points were 7.17±2.42 mm before and 1.41±0.42 mm after self-registration. In all our experiments, final errors with our method were less than the voxel size.

The results of FOD driven registrations were also compared to those obtained from T1W and DT fractional anisotropy (FA) data using an identical ITK based FFD technique. For each technique, data from the above mentioned eight subjects were registered to a representative template. Registrations were performed on 3 CP spacing levels (FA & FOD, 24,16, and 8mm with 30,20, and 10 iterations, and T1W 20,10, and 5mm with 50, 40, and 30 iterations). T1W images were rigidly co-registered to b=0 images. All transforms were composed into a single deformation, applied to FOD and FA volumes, averaged, and visually assessed for sharpness indicative of correct spatial alignment. As illustrated in figure 1b, the FOD registration results are sharper than the FA (fig 1c) and T1W (fig 1d) results.

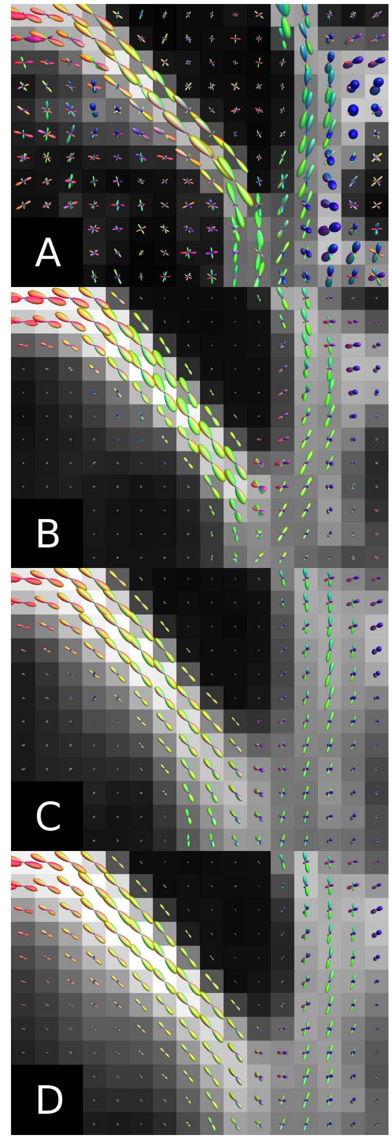

Figure 1. An axial view of the Splenium of the Corpus Callosum, average of 8 subjects registered to a single template. FODs are overlayed on an FA map. **A)** Template **B**) FOD Registered **C)** FA Registered **D)** T1W Registered

Discussion and Conclusion

A novel DWI registration technique has been demonstrated to exploit intra-voxel fibre orientation information provided by FODs to enable improved accuracy in spatial normalisation. More accurate spatial normalisation will benefit group based clinical and neuroscientific studies, as well as automatic atlas based anatomical labelling, which are increasingly important application areas for diffusion MRI.

References

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