Fiber-Tractography in Human Brain Using Diffusion Tensor MRI (DT-MRI)

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Purpose
To elucidate fiber-tract trajectories in human brain white matter using in vivo diffusion tensor MRI (DT-MRI) data, to provide a framework for testing tract following schemes, and to consider artifacts inherent in them.

Introduction
The elucidation of fiber tract trajectories in brain and in other fibrous tissues should provide valuable information to assess fiber connectivity and continuity, and provide insight into the natural history of normal and abnormal development and of related diseases. Since the tractography algorithm used here was first proposed (1), several novel methods have been advanced to follow fiber tract trajectories using measured DT-MRI data (2-5). The work here is relevant to these studies as we show several artifacts that have not been considered to date, and suggest means of remediating them.

Theory
DT-MRI (6) provides a means to generate fiber-tract trajectories. In each voxel (containing coherent fibrous tissue), the local fiber tract direction is parallel to the eigenvector, \( \mathbf{e}_1 \), associated with the largest eigenvalue, \( \lambda_1 \), of the diffusion tensor, \( \mathbf{D} \) (6). If we represent the fiber tract trajectory as a vector, \( \mathbf{r}(s) \), parameterized by arc length, \( s \), then the spatial evolution of \( \mathbf{r}(s) \) is described by a vector differential equation (7) shown on the left side below:

\[
\frac{d\mathbf{r}(s)}{ds} = \mathbf{t}(s); \quad \mathbf{t}(s) = \mathbf{e}_1(\mathbf{r}(s))
\]

When using DT-MRI data we equate \( \mathbf{t}(s) \), the unit vector tangent to \( \mathbf{r}(s) \), and \( \mathbf{e}_1(\mathbf{r}(s)) \), the normalized eigenvector of \( \mathbf{D}(\mathbf{r}(s)) \) associated with its largest eigenvalue, \( \lambda_1(\mathbf{r}(s)) \) (1).

However, the \( \mathbf{e}_1 \)-field that we measure using DT-MRI is a noisy, discrete, voxel-averaged statistical estimate of the true direction field. Thus, attempting to construct a continuous fiber tract trajectory using measured \( \mathbf{e}_1 \) data is problematic. To address this issue, we developed robust, computationally efficient algorithms for \( \mathbf{D} \)-field processing (8), that we use to generate a continuous, smooth approximation, \( \mathbf{D}(\mathbf{x}) \), to the measured DT-MRI data at each position \( \mathbf{x} \) within the imaging volume. We use \( \mathbf{D}(\mathbf{x}) \) to generate a continuous \( \mathbf{e}_1 \)-field, from which fiber tract trajectories can be calculated using the equations above.

Methods
The Runge-Kutta method (9) was used to solve the differential equations above for \( \mathbf{r}(s) \). To test algorithm performance, a family of analytically derived \( \mathbf{D} \)-fields were synthesized and digitally resampled (see (1)). Background noise was also added using Monte Carlo methods (10).

DT-MRI was performed on normal volunteers in a 1.5T Signa scanner equipped with whole body gradient coils using imaging parameters given in (11).

Results
No deviations from the known tract trajectory were observed when using synthetically generated noiseless \( \mathbf{D} \)-fields (1) except at singularities in the fiber-tract direction field, i.e., regions where tracts cross, “kiss”, branch, merge or splay. The ability to follow fiber tracts is diminished as SNR decreases. This problem is ameliorated by first smoothing these noisy \( \mathbf{D} \)-fields before applying the tract following algorithm.

This tract following schema was then applied to in vivo DT-MRI data of human brain. Plausible fiber trajectories were consistently found in thick white matter structures, such as in the corpus callosum. In other thinner coherent pathways, such as in the pyramidal tract, the algorithm’s performance is less robust, depending more upon parameters, such as the size of the smoothing window.

Fig 1: Fiber-tract trajectory of corpus callosum in human brain

Discussion and Concluding Remarks
Using these approaches, we can generate anatomically plausible fiber tract trajectories in large coherently oriented white matter tracts, using in vivo DT-MRI data. Singularities in the fiber direction field, causing “powder averaging” of the \( \mathbf{D} \)-field (10), and background noise in DWIs introduce distinct artifacts in tract following applications, which must be considered in clinical studies relating developmental, neurological, or psychiatric disorders to structural alterations in white matter fiber tract architecture.

References