Measurement of Intrinsic Fiber Diffusivity Using Spherical Deconvolution of High Angular Resolution Diffusion Data

H-K. Jeong[1,2], and A. W. Anderson[1,2]

[1]Vanderbilt University Institute of Imaging Science, Nashville, TN, United States, [2]Biomedical Engineering, Vanderbilt University, Nashville, TN, United States

INTRODUCTION High angular resolution diffusion imaging (HARDI) and other methods map the architecture of white matter fibers by measuring the water diffusion within each voxel of MR images (1, 2). However, while they can identify multiple fiber directions, they do not provide estimates of the intrinsic diffusion properties of any of the fibers. The present study describes a method to resolve and estimate the diffusivities of each fiber within a voxel using the FORECAST model (3).

METHODS The diffusion weighted (DW) signal can be expressed as the spherical convolution of $P(\theta',\phi')$, the fiber angular distribution function (FAD), with the response to a single fiber. To resolve individual fiber's radial diffusivity ($\lambda_{\perp k}$) for the multiple fiber case, the error function $E(\hat{\boldsymbol{\eta}}) = \sum\sum\left(S(\mathbf{r}_k,b_j,\boldsymbol{\eta}) - \hat{S}(\mathbf{r}_k,b_j,\hat{\boldsymbol{\eta}})\right)^2$ is to be minimized, where $\eta(b_j,\lambda_{\perp k}) = b_j(\bar{\lambda}-\lambda_{\perp k})$, $\mathbf{r}_k$ is k-th diffusion gradient vector, b_j is j-th b-value and $\bar{\lambda}$ is mean diffusivity. The model signal $\hat{\mathbf{S}}$ is generated for the error function using the spherical harmonic transform of the observed signal (s_{lm}) and FAD (p_{lm}) for each trial value of $\boldsymbol{\eta}$ via the simple matrix representation $\hat{\mathbf{S}} = S_0 \cdot \mathbf{Z} \cdot \mathbf{C} \cdot \mathbf{p}_{lm}$, where $\mathbf{C}$ holds the Legendre series coefficients of the DW signal kernel function and $\mathbf{Z}$ is a matrix of spherical harmonics.

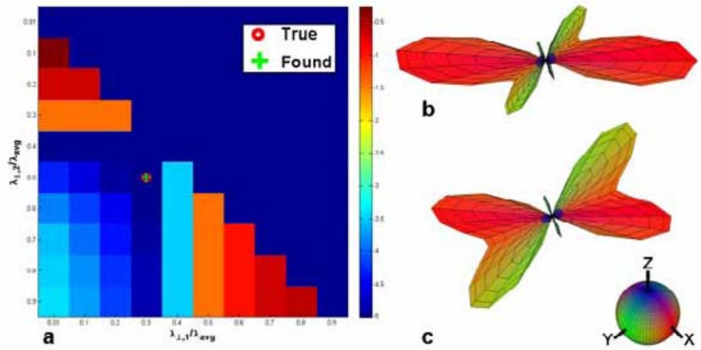

Fig. 1. True radial diffusivity for each fiber is found on a discrete search grid of the error function in $\lambda_{\perp k}/\bar{\lambda}$ unit (a). The volume fraction of each fiber bundle is estimated by the amplitude of the FAD, shown here for the case of a single response function, calculated from $\lambda_{\perp s}$ (b) and two response functions, calculated from $\lambda_{\perp 1}$ and $\lambda_{\perp 2}$ (c).

A numerical simulation of two crossing fibers with equal volume fractions was developed to test the ability of this model to recover the true radial diffusivity and volume fraction for individual fibers. The radial diffusivity of the individual fibers ($\lambda_{\perp 1}$ and $\lambda_{\perp 2}$) can be assumed to satisfy $\lambda_{\perp 1} < \lambda_{\perp 2}$, $0 < \lambda_{\perp 1} \le \lambda_{\perp s}$ and $\lambda_{\perp s} < \lambda_{\perp 2} < \bar{\lambda}$, where $\lambda_{\perp s}$ is the radial diffusivity for a single fiber fit. Four in-vivo data sets are acquired for the analysis using a Philips 3T Achieva with SENSE head coil, b-values {0, 1000, 2000} sec/mm^2, 2.5 mm^3 isotropic voxels, 96 x 96 scan size, TR 9000 and TE 55 msec.

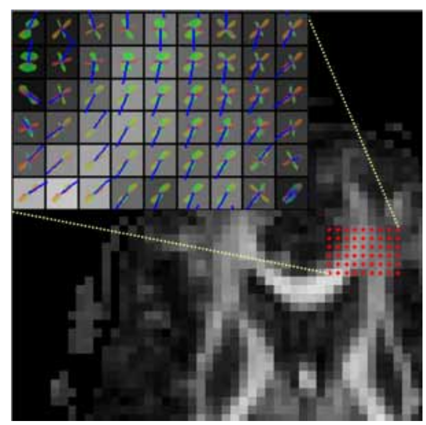

Fig. 2. FAD is shown on the FA map for an ROI (red dots) near the junction of corpus callosum and anterior region of corona radiata. The blue bar for each voxel shows the principal eigenvector of the diffusion tensor.

RESULTS Simulation results are shown in Fig.1. Each fiber's true radial diffusivity is found over search grid as shown in Fig.1a. The volume fractions for individual fiber bundles are shown assuming a single value of $\lambda_{\perp s}$ (Fig.1b) and from the fit for both $\lambda_{\perp 1}$ and $\lambda_{\perp 2}$ (Fig. 1c). This result shows that this model correctly estimates each fiber's perpendicular diffusivity and volume fraction which is not possible with a single convolution kernel (Fig.1b). The FA and FAD map of an ROI for the in-vivo data are shown in Fig.2. The result for the estimation of the lower and higher radial diffusivities are shown in Fig. 3a and 3b.

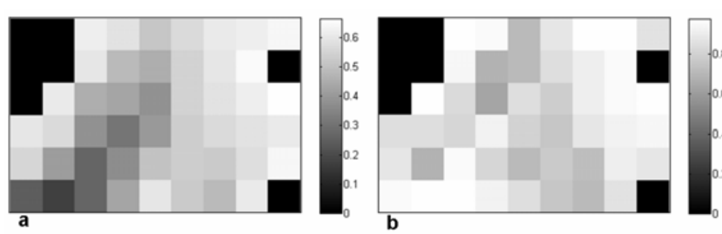

Fig. 3. Radial diffusivities for the first (lower) (a) and the second (higher) fiber (b) are shown as ratios, $\lambda_{\perp k}/\bar{\lambda}$, unit for the voxels with FA > 0.22.

DISCUSSION This study assumes that each fiber compartment has axially symmetric diffusion, the same relaxation rate, negligible exchange between compartments and constant mean diffusivity (3, 4). Because the error function is only weakly dependent of $\{\lambda_{\perp k}\}$ errors, increasing the signal to noise ratio is important for reliable estimates in the presence of image noise. In this study we used four acquisitions with bootstrap sampling for the analysis. This study implies, however, that this approach may help identify fiber anisotropy and volume fraction and tracking of fiber bundles within a single voxel.

REFERENCES 1. Tuch DS, et al. Proc. ISMRM 7, 321. 2. Wedeen VJ, et al. Magn Reson Med 2005;54:1377-1386. 3. Anderson AW. Magn Reson Med 2005;54:1194-1206. 4. Pierpaoli C, et al. Radiology 1996;201:637-648.