Tensor Distribution Function in Multiple Shell High Angular Resolution Diffusion Imaging

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Introduction: DTI reveals white matter microstructure and fiber pathways in the living brain by examining the 3D diffusion profile of water molecules in brain tissue. Even so, DTI-derived measures will be incorrect where fibers cross or mix, as the single tensor model cannot resolve these more complicated white matter configurations. This issue can be addressed using the model-free diffusion spectrum imaging (DSI), which exploits the direct Fourier inversion of the diffusion signal. This technique is time intensive, as it measures the signal on a 3D Cartesian lattice [1]. An alternative approach based on sampling only on one or multiple spherical shells in q-space has been proposed, referred to as high angular resolution diffusion imaging[2]. Each spherical shell, being a 2D manifold with a specific b value [3], includes a number of measurement points which grows quadratically with the desired angular resolution, as opposed to cubically with the spatial resolution in the entire 3D lattice of DSI.

Methods: In this study, we illustrated how to manipulate multiple shell HARDI data using Tensor Distribution Function (TDF). [4] A healthy human brain was scanned using a singly refocused 2D single shot spin echo EPI sequence in 7T field strength. Image parameters were: FOV: 192×192 mm^2 (matrix: 196×96) to yield a spatial resolution of 2×2×2 mm^3, TR/TE 4800/57 msec., acceleration factor (GRAPPA) of 2 and 6/8 partial Fourier were used along the phase encode direction. Diffusion-weighted images were acquired at three b-values of 1000, 2000 and 3000 s/mm^2 with 256 directions, along with 31 baseline images. EPI echo spacing was 0.57 msec. with a bandwidth of 2895 Hz/Px. We model the HARDI signal as a unit-mass probability density on the 6D manifold of symmetric positive definite tensors, yielding a continuous mixture of tensors, at each point in the brain, which is the key of TDF theory [4]. The TDF can model fiber crossing and non-Gaussian diffusion. From the TDF, one can derive analytic formulae for the water displacement probability function, orientation distribution function (ODF), tensor orientation distribution function (TOD), and their corresponding anisotropy measures. Here we further develop the TDF framework for multiple shell. (Figure 1)

Results: We plot ODF for different shells and also multiple shell for the axial view. (Figure 2) The color in the ODF plot indicates the directions: red corresponds to medial-lateral, green to anterior–posterior, and blue to superior–inferior orientation. From the figure, we can see that low b shell data has a noisy ODF plot, while high b shell data has partial information loss in the fiber crossing region. The multiple shell ODF can address these two disadvantages by integrating all information from different shells.

Conclusions: The tensor distribution function is a powerful signal reconstruction method that can resolve intravoxel fiber crossing in multiple shells HARDI, which can integrate greater information from different shells in compared with single shell.

References: 1. Wedeen VJ, Hagmann P, Tseng WI, Reese TG, Weisskoff RM. Mapping complex tissue architecture with diffusion spectrum magnetic resonance imaging. Magnetic Resonance in Medicine. 2005;54(6):1377–1386.
2. Tuch DS, Reese TG, Wiegell MR, Makris N, Belliveau JW, Wedeen VJ. High angular resolution diffusion imaging reveals intravoxel white matter fiber heterogeneity. Magnetic Resonance in Medicine. 2002;48(4):577–582.

3. LeBihan D. 1990. Magnetic resonance imaging of perfusion. Magnetic Resonance in Medicine 14(2): 283-292.
4. Leow, AD. (2009), 'The tensor distribution function', Magnetic Resonance in Medicine, vol. 61, no. 1, pp. 205-214.

Shell 1	b_1					
	q_{11}	q_{12}	...	q_{1j}	...	q_{1N}
Shell 2	b_2					
	q_{21}	q_{22}	...	q_{2j}	...	q_{2N}
.....						
Shell i	b_i					
	q_{i1}	q_{i2}	...	q_{ij}	...	q_{iN}
.....						
Shell M	b_M					
	q_{M1}	q_{M1}	...	q_{Mj}	...	q_{M1}
Multiple Shell	b_1		b_2		...	b_M
	$q_{11}\ q_{12}$ .....q_{1N}		$q_{21}\ q_{22}$ .....q_{2N}			$q_{M1}\ q_{M2}$ .....q_{MN}

b_i is the instrumental scaling factor, which is specific corresponding to each shell.
q_{ij} is the diffusion-sensitized gradient vector for the jth direction in the ith shell

Single Shell TDF	$S_{calculated}\ (q_i) = \int_{D \in \mathbb{D}} P(D)\exp(-bq_i{}^T D q_i)dD$ $P^* = argmin_p \sum_{i=1}^{N} (S_{obs}(q_i) - S_{calculated}\ (q_i))^2$
Multiple Shell TDF	$S_{calculated}\ (q_{ij}) = \int_{D \in \overline{\mathbb{D}}} P(D)\exp(-b_i q_{ij}{}^t D q_{ij})dD$ $P^* = argmin_p \sum_{i=1}^{M}\sum_{j=1}^{N} (S_{obs}(q_{ij}) - S_{calculated}\ (q_{ij}))^2$
$ODF(\tilde{x}) = C\int_{D \in \overline{D}} P(D)(\det(D)\,\tilde{x}^t D^{-1}\tilde{x})^{-\frac{1}{2}}dD$	

Figure 1

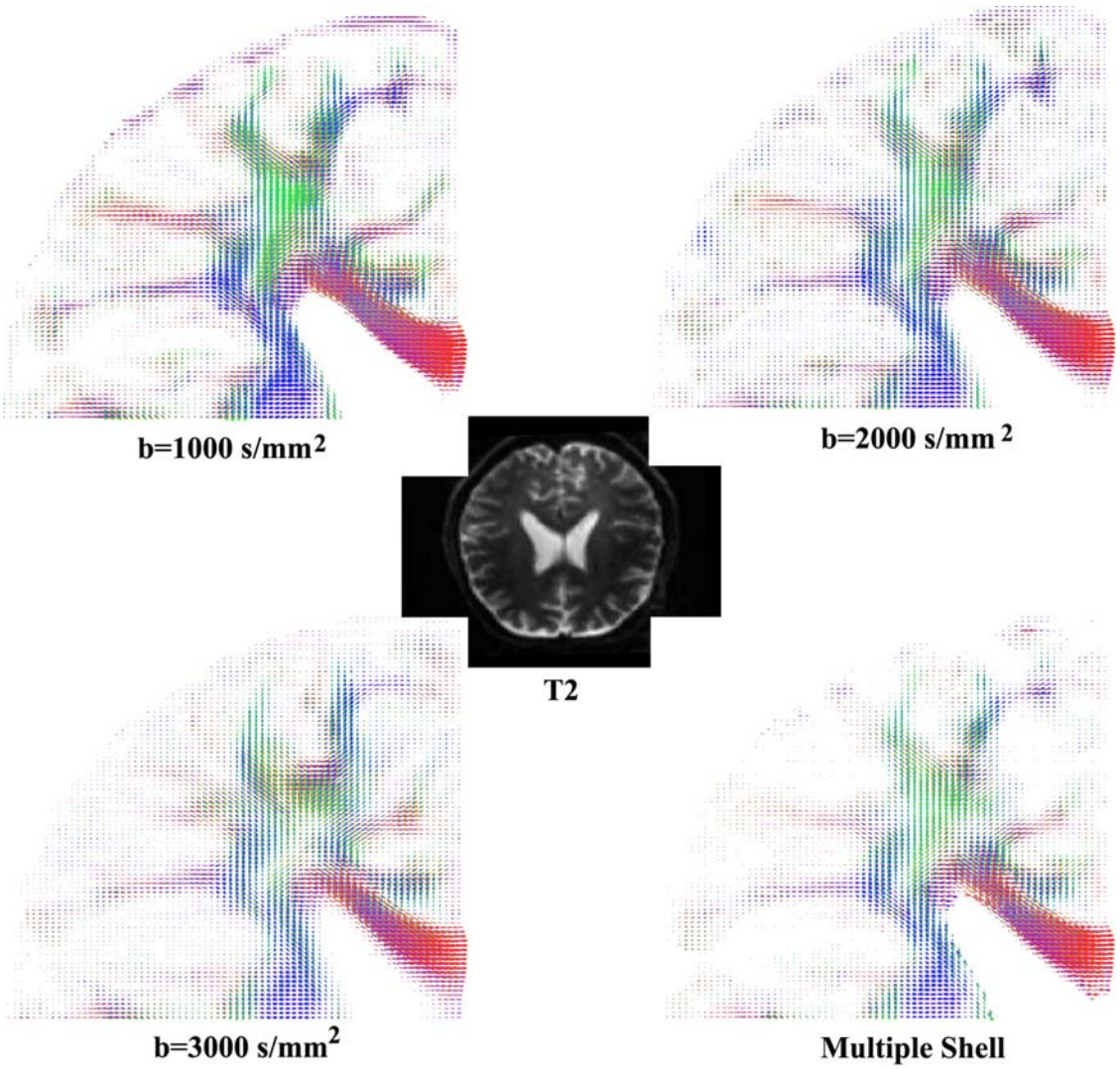

Figure 2. ODF plot for different shells and multiple shell. From this figure, we can see that low b value (e.g. 1000) can cause noisy ODF plot, while the high b value (e.g. 3000) lead to missing information in ODF, the multiple shell can avoid both of the disadvantages and combine greater information from different shells.