In vivo investigations of accuracy and precision of fiber orientations in crossing fibers in spherical deconvolution-based HARDI methods

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Target audience: Researchers interested in optimizing High-Angular Resolution Diffusion Imaging acquisitions.

Introduction: Diffusion MRI methods based on high-angular resolution diffusion imaging (HARDI) acquisitions provide a good trade-off between acquisition time and accuracy of the estimated diffusion or fiber orientation distribution function (dODF/fODF) [1-3]. Recently, an extensive evaluation was done on the effect of the number of diffusion-weighting (DW) gradient orientations and the b-value on the accuracy of a spherical harmonics representation of the DW signal of a single fiber population [3]. A b-value of 3000 s/mm^2 was shown to be optimal to describe the angular profile of the signal [3]. Both the maximum spherical harmonic order detected and the accuracy of the fit are strongly dependent on the effective angular SNR (i.e., the SNR and the number of DWIs) and b-value. This work presents an overview of how these factors affect fODF estimation in crossing fiber voxels in vivo, investigating the accuracy and precision of the reconstructed fODF peaks. Specifically, we intend to determine if there are situations were low b-value and a high number of DWIs is equal to or better than a high b-value and a relatively low number of DWIs (e.g., the performance of CSD is at b=1000 s/mm^2 with 200 DWIs equals that at b=3000 s/mm^2 with only 100 DWIs), and what would be optimal in terms of scan time.

Methods: Acquisition: a DWI dataset of a 24 year-old healthy subject was acquired on a 3.0 T Philips Achieva, with three shells at b-values of 1000, 2000, and 3000 s/mm^2 with 250 gradient directions in each shell, and 75 b=0 images. The images from the three shells were acquired across different sessions (similar to the repeated sessions in [4]). Acquisition details: matrix 96×96; FOV 240×240; 2.5 mm slices for an isotropic voxel size of 2.5 mm^3 [5]; SENSE=2.5; TE/TR: 100/7500 ms; SNR of b=0 images about 32 in the WM. Test datasets: For all shells, ten subsets were created for sets with the following number of gradient directions: {50, 60, 70, 80, 90, 100, 125, 150, 175, 200, 225}, as described in [6]. fODF estimation: For all subsets, fODFs were estimated using CSD (*Lmax* 6 or 8) in ExploreDTI, where the response function was estimated recursively [2,7-9]. Analysis: fODF peak orientations were extracted to investigate the accuracy and precision for each combination of b-value, *Lmax*, and number of gradient directions (**Fig. 1**). Two voxels were investigated: one voxel with two distinct fiber populations (bottom indicated voxel in **Fig. 1a**); and one with three (top indicated voxel in **Fig. 1a**). A "silver standard" peak orientation was obtained for each fiber population from the b=3000 data modeled with *Lmax*=8 by creating 100 subsets of the 225 directions and averaging the peak orientations over these 100 subsets. For each of the other combinations of b-value, *Lmax*, and number of directions, mean peaks were obtained in a similar fashion, averaging over the 10 subsets. The mean angular deviation of all these mean peaks was calculated by taking the angle with respect to the corresponding peak from the "silver standard", as a measure of accuracy. The angular deviation is taken as a measure of precision, and calculated as the standard deviation over peak orientations of the 10 subsets per setting.

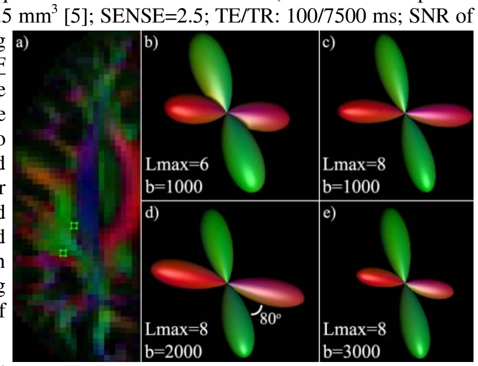
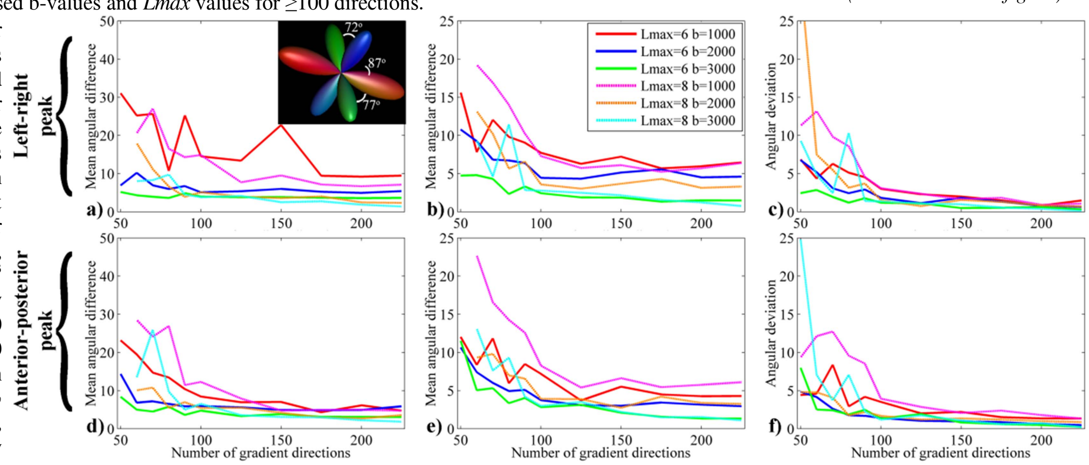


Fig. 1: *The two voxels investigated are shown in **a)** by the green boxes. **b)-e)** show fODFs for one subsets with 100 gradient directions (the bottom indicated voxel), from datasets with different b-values and* Lmax *values (as denoted in the figure).*

Results: Example fODFs for a two-fiber voxel are shown in **Fig. 1** for different b-values and *Lmax* settings. **Fig. 2** shows the mean angular difference and the angular deviation for each set of parameters. For all the fODF peaks and settings investigated, the mean angular difference reaches a stable value with increasing number of gradient directions. In terms of accuracy, the increase in effective angular resolution with b-value means that less gradient directions are required to reach this high accuracy. The angular deviation (**Fig. 2c,f**) is very stable and very low, indicating a very good precision for all used b-values and *Lmax* values for ≥100 directions.

Discussion: Confirming the single-fiber voxels results from [3], higher b-values provide high levels of precision and accuracy, even for relatively low number of gradient orientations (50-100). On the whole, more DWIs at lower b-values performed to a similar level of precision as high b-values with fewer DWIs. At standard clinical 3 T MR systems, higher b-values require longer DW gradients, increasing TE and thus TR. At a 2.5 mm^3 resolution, this increase in TE is roughly 10 ms to increase the b-value from 1000 to 2000 and another 10 ms for b=3000 (at G_{MAX} 30-40 mT/m). This results in an increase in TR of roughly 8 and 16% from b=1000 to b=2000 and b=3000, respectively. For any given accuracy (**Fig. 2**), increasing the b-value is more scan-time-efficient than increasing the number of DWIs. Future work will focus on 1) investigating the effect of TE prolongation at higher b-values on the SNR, and 2) varying the SNR to determine if these results generalize to more imaging settings, e.g., higher spatial resolutions where the SNR is typically lower.



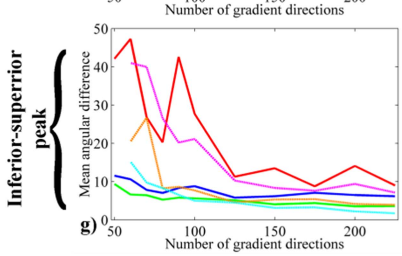


Fig. 2: *Mean angular difference (in degrees) for a voxel with three (**a,d,g**) and two (**b,e**) fiber populations; and the angular deviation (in degrees) for the voxel with two populations (**c,f**). Inset in **a)** shows example fODF (same setting as **Fig. 1d**). For all b-values, about half of the subsets containing 50 gradient directions could not estimate the fODF at* Lmax*=8, where the optimal subsets of 50 directions were apparently insufficient or insufficiently uncorrelated to estimate the required 45 parameters. These are not shown for the mean angular difference plots (**a,b,d,e,g**).*

References: [1] Tuch, MRM 2004;52:1358-72; [2] Tournier et al., NeuroImage 2007;35:1459-72; [3] Tournier et al., NMR Biomed 2013, in press; [4] Jeurissen et al., HBM 2012, in press; [5] Vos et al., ISMRM 2011 p1945; [6] Vos et al., ISMRM 2013 p842; [7] Leemans et al., ISMRM 2009 p3536; [8] Jeurissen et al., HBM 2011;32:461-79; [9] Tax et al., NeuroImage 2013, in press.