The effect of the acquisition parameters for High Angular Resolution Diffusion Imaging

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INTRODUCTION: One of the recent challenges in diffusion imaging is to find acquisition schemes and analysis approaches that can represent non-Gaussian diffusion profiles in a clinically feasible measurement time. In this work we investigate the effect of the b value and the number of gradient vector directions on the HARDI models: Q-ball imaging and the Diffusion Orientation Transform (DOT) in a structured way using computational simulations, hardware crossing-fiber diffusion phantoms, and in-vivo brain scans. We quantify the angular resolution of the reconstructed probability profiles under different acquisition schemes. We vary the b value and number of gradient directions to investigate their effects and interaction. Knowledge of the suitable acquisition parameters in clinical setting, for different HARDI diffusion models, where still satisfying results are obtained given the high noise impact on the data, is of significant importance in this emerging area.

METHODS: We implemented the analytical Q-ball imaging [1] and the parametric and nonparametric DOT [2] in C++. As the results of DOT depend on the optimal radius of the sphere that defines the probability function R0 and the effective diffusion time t, we experimentally found R0 and extracted t from the imaging parameters as t = Δ/δ. In our simulations we found the following values of R0 = (22, 24, 26, 28, 30) μm to give minimal angular error for the DOT. We varied the order of the Spherical Harmonics l between 4 and 8. As a preprocessing step for the noisy data sets, we included Laplace-Beltrami (LB) smoothing on the signal, for both of the methods with λ = 0.006 as in [3]. We assume that the fiber directions are given by the local maxima of the normalized [0,1] ODF/probability profile where the function surpasses a threshold (0.5 like in [1]).

DATA: Synthetic data of 2-fiber crossings under 40°, 45°, 50°, 55°, 60° and 90° were generated in Mathematica by simulating the DW-MR signal attenuation from molecules restricted inside a cylinder [4]. Gaussian noise was added to the real and complex part of the signal with realistic SNR corresponding to the used b values (15.3, 14.1, 13.3, 12.0 and 11.9 for 1000, 1500, 2000, 3000 and 4000 s/mm² respectively). DW-MRI hardware phantoms were constructed of 2-fiber crossings under 30°, 50°, and 65° [5]. DW-MRI acquisition was performed on subject VP (25 yrs, female) using a twice refocused SE-EPI sequence on a Siemens Allegra 3T (Siemens, Erlangen). FOV 208 × 208 mm, voxel size 2.0 × 2.0 × 2.0 mm. 10 horizontal slices were positioned through the body of the corpus callosum and centrum semiovale. Data sets were acquired with #vols(#dirs): 132(120) 106(96), 80(72), 54(48) directions, each at b values of 1000, 1500, 2000, 3000, 4000 s/mm². The phantom was scanned using exactly the same DW-MRI protocol as the human subject (the 54 directions scheme was omitted). We implemented the analytical Q-ball imaging [1] and the parametric and nonparametric DOT [2] in C++. As the results of DOT depend on the optimal radius of the sphere that defines the probability function R0 and the effective diffusion time t, we experimentally found R0 and extracted t from the imaging parameters as t = Δ/δ. In our simulations we found the following values of R0 = (22, 24, 26, 28, 30) μm to give minimal angular error for the DOT. We varied the order of the Spherical Harmonics l between 4 and 8. As a preprocessing step for the noisy data sets, we included Laplace-Beltrami (LB) smoothing on the signal, for both of the methods with λ = 0.006 as in [3]. We assume that the fiber directions are given by the local maxima of the normalized [0,1] ODF/probability profile where the function surpasses a threshold (0.5 like in [1]).

RESULTS: In noiseless simulation (Fig. 1), DOT gives the same results as Q-ball but at lower b value and less gradient directions. On Fig. 2 and Fig. 3, results of the noisy synthetic and hardware phantom data analysis are given. The angular error is plotted for the smallest angle found within our criteria (the angular error and the standard deviation to be smaller than a liberate threshold of 20), for each combination of b value and direction table (given on the x-axis), both for DOT (upper panel) and Q-ball (lower panel). On Fig.2 we observe that DOT recovers in most of the cases smaller angles than Q-ball for the same se

CONCLUSION: Our results suggest that over a wide range of acquisition schemes DOT can outperform Q-ball in realistic simulations, hardware diffusion phantom data and in-vivo brain data. However tuning the optimal R0 for the DOT, is not trivial in in-vivo data sets, and by simplicity and speed of calculations of the reconstructed ODFs, Q-ball still has a significant advantage. In this work we also explored the effect of LB smoothing on the signal for DOT. The signal attenuation in DOT is assumed to be mono-exponential for b <2000 s/mm², which can be seen as a smoothing in q-space already. Therefore LB smoothing slightly improves the properties of DOT in some specific cases (Fig. 5). Knowledge of suitable acquisition schemes in a clinical setting, that can correctly estimate non-Gaussian diffusion profiles for HARDI, can improve the utility of DW-MRI for the diagnosis of white matter diseases and presurgical planning.