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A new multi-directional fiber model for low angular resolution diffusion imaging

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Introduction: Diffusion MRI (dMRI) permits to infer the ensemble average propagator (EAP) from a set of diffusion-weighted (DW) images acquired from \( n_b \) gradient directions and \( n_b \) b-values. In the context of clinical brain imaging, dMRI sequences seldom exceed 10 minutes acquisition, with \( n_b \leq 30 \) and only one b-value. The EAP is then inferred from the resulting low angular resolution diffusion (LARD) images by assuming a Gaussian diffusion profile [1]. In research context, higher angular resolution samplings (\( n_b \geq 60 \) and \( n_b \geq 1 \)) [2,3,4] have revealed a non-Gaussian diffusion profile in the white matter when fibers cross. To account for that effect, we propose a non-Gaussian parametric modeling of the EAP, the estimation of which can be accurately performed from LARD images obtained in clinical context.

Theory: In each voxel, the EAP is modeled as a mixture. Each probability density function (pdf) in the mixture characterizes the diffusion along some fiber orientation (FO) \( \mu \), \( ||\mu||=1 \), and is in turn modeled as a mixture of two equally weighted pdfs that account for the diffusion along directions \( \mu \) and \(-\mu\) respectively. The diffusion pdf along direction \( \mu \) is given by the convolution of a von Mises & Fisher pdf on the sphere of radius \( R>0 \) (mean covered distance), with mean direction \( \mu \) and concentration (around \( \mu \)) parameter \( \kappa \geq 0 \), and a centered 3D Gaussian pdf with covariance matrix \( D=R^2(1+\kappa\mu\mu^T)/(\kappa+1) \) (cylindrical shape). Crossing fibers are consequently characterized by 8 parameters. The Fourier transform of the EAP is analytically derived as a function of the parameters of the model and yields the theoretical DW intensity [5]. The estimation of these parameters is then performed by a least squares fitting of the observed DW intensities to the theoretical ones.

Methods: An evaluation of the crossing angle resolution (CAR) of the model was first performed using synthetic data on a single voxel. These data were generated as in [6] for different configurations of the two FOs with \( b = 1500 s/mm^2 \) and \( n_b = 15, 30, 41, 64 \) and 200. The resulting data sets were then corrupted with increasing Rician noise and, for each noise level \( \sigma \), 100 samples were synthetized. For a given \( n_b \) and \( \sigma \), the CAR was computed as the 95% confidence angle between the two estimated FOs in situations where the real FOs are collinear. A healthy adult male was scanned on a 3T Achieva Philips MRI Scanner with a 8-ch head coil, TR/TE/\( \tau = 10000/64/22.1\)ms, \( b=8000 s/mm^2 \), \( n_b = 15 \) and 2x2x2mm\(^3\) voxels. This set of DW images represents a typical case of LARD images with low spatial resolution from which our model of the EAP was estimated.

Results: Figure on the left shows the CAR of the model for increasing signal-to-noise ratios (SNR). Each curve corresponds to a given \( n_b \). For low SNRs, increasing \( n_b \) does not significantly improve the CAR. For typical clinical values of SNR = 20dB and \( n_b = 30 \), the corresponding CAR of 30° outperforms the CAR obtained in Q-Ball Imaging [7], i.e. around 60° for higher angular resolution (\( n_b = 81 \)) [8]. Figure on the bottom shows an extremity of the corpus callosum known to contain crossing fibers (the height of the cones is proportional to \( R^2 \) while the radius is proportional to \( 1/(\kappa+1) \)). Fiber crossings seem to be accurately estimated despite the low angular and spatial resolutions.

Discussion: This model enables crossing fibers to be theoretically estimated from only 8 DW images. In particular, this model allows for the retrospective study of DW data sets acquired over the past years. For a complete applicability in clinics, one could wonder whether maps akin to the fractional anisotropy (FA) and mean diffusivity (MD) maps [9] can be provided with this model. For a given FO, based on the Gaussian part of our model and by analogy with DTI, we propose \( FA = \kappa((\kappa+1)^2-2)^{-1/2} \) and \( MD = (1+\kappa/3) R^2/(1+\kappa) \).