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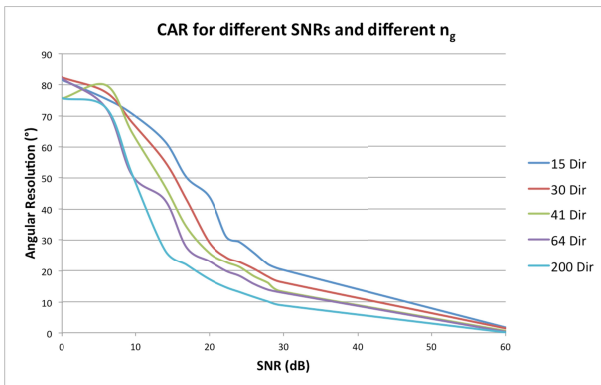
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Introduction: Diffusion MRI (dMRI) permits to infer the ensemble average propagator (EAP) from a set of diffusion-weighted (DW) images acquired from n_g gradient directions and n_b b-values. In the context of clinical brain imaging, dMRI sequences seldom exceed 10 minutes acquisition, with $n_g \leq 30$ 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 ($n_g \geq 60$ 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) $\pm \mu$, $||\mu||=1$, and is in turn modeled as a mixture of two equally weighted pdfs that account for the diffusion along directions μ and $-\mu$ respectively. The diffusion pdf along direction μ is given by the convolution of a von Mises & Fisher pdf on the sphere of radius $R>0$ (mean covered distance), with mean direction μ and concentration (around μ) parameter $\kappa \geq 0$, and a centered 3D Gaussian pdf with covariance matrix $D=R^2 (I_3 + \kappa \mu \mu^T)/(\kappa+1)$ (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 $b = 1500 \text{ s/mm}^2$ and $n_g=15, 30, 41, 64$ and 200. The resulting data sets were then corrupted with increasing Rician noise and, for each noise level σ , 100 samples were synthesized. For a given n_g and σ , 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, $TR/TE/\tau = 10000/64/22.1 \text{ ms}$, $b=800 \text{ s/mm}^2$, $n_g=15$ and $2 \times 2 \times 2 \text{ mm}^3$ 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_g . For low SNRs, increasing n_g does not significantly improve the CAR. For typical clinical values of $\text{SNR} = 20 \text{ dB}$ and $n_g=30$, the corresponding CAR of 30° outperforms the CAR obtained in Q-Ball Imaging [7], i.e. around 60° for higher angular resolution ($n_g=81$) [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 $MD = (1+\kappa/3) R^2/(1+\kappa)$.

References: [1] Basser et al., Biophys. J. 66/1, 259-67, 1994 [2] Wedeen et al., MRM 54, 1377-86, 2005 [3] Descoteaux et al., MEDIA, 1-19, 2010 [4] Tuch et al., MRM 48, 577-82, 2002 [5] P.T. Callaghan, 1991 [6] Barmoutis et al., IPMI 21, 338-49, 2009 [7] D. Tuch, MRM 52/6, 1358-72, 2004 [8] Descoteaux et al., MRM 58, 497-510, 2007 [9] Basser et al., J. Magn. Reson. B-111, 209-19, 1996.

