



**HAL**  
open science

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

Aymeric Stamm, Patrick Pérez, Christian Barillot

► **To cite this version:**

Aymeric Stamm, Patrick Pérez, Christian Barillot. A new multi-directional fiber model for low angular resolution diffusion imaging. International Society for Magnetic Resonance in Medicine, May 2012, Australia. pp.908. inserm-00858206

**HAL Id: inserm-00858206**

**<https://inserm.hal.science/inserm-00858206>**

Submitted on 4 Sep 2013

**HAL** is a multi-disciplinary open access archive for the deposit and dissemination of scientific research documents, whether they are published or not. The documents may come from teaching and research institutions in France or abroad, or from public or private research centers.

L'archive ouverte pluridisciplinaire **HAL**, est destinée au dépôt et à la diffusion de documents scientifiques de niveau recherche, publiés ou non, émanant des établissements d'enseignement et de recherche français ou étrangers, des laboratoires publics ou privés.

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

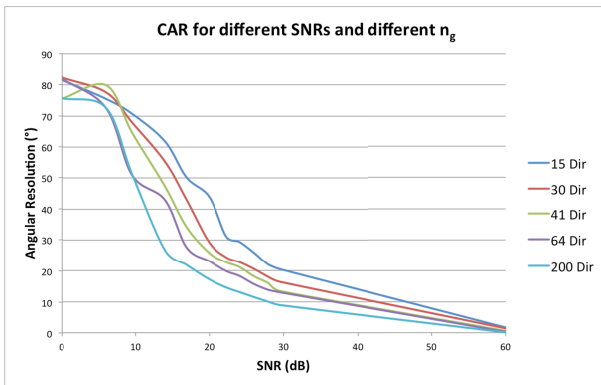
Aymeric Stamm<sup>1</sup>, Patrick Pérez<sup>2</sup>, and Christian Barillot<sup>1</sup>

<sup>1</sup>Visages INSERM/INRIA U746, IRISA - UMR CNRS 6074, Rennes, France, <sup>2</sup>Technicolor, Rennes, France

**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  $\mu$  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  $D = R^2 (I_3 + \kappa \mu \mu^T) / (\kappa + 1)$  (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  $\sigma$ , 100 samples were synthesized. For a given  $n_g$  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,  $TR/TE/\tau = 10000/64/22.1 \text{ ms}$ ,  $b = 800 \text{ s/mm}^2$ ,  $n_g = 15$  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  $30^\circ$  outperforms the CAR obtained in Q-Ball Imaging [7], i.e. around  $60^\circ$  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] Barmpoutis 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.

