# A Modified Damped Richardson-Lucy Algorithm to Improve the Estimation of Fiber Orientations in Spherical Deconvolution

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**Introduction:**. Standard spherical deconvolution approaches assume that HARDI signal can be modeled as a convolution between a common fiber response and a fiber orientation distribution function [1][2]. Recent spherical deconvolution methods improved the quality of the results introducing the non negative constraint of the solution [3][4][5]. It can be verified, however, that instability problems, such as spurious fiber orientations, could be related not only to noise robustness but also to signal contributes, such as from isotropic tissues, that are not properly taken into account. To develop new and more advanced fiber tracking techniques it is essential that fiber orientations are meaningful not only in main white matter (WM) fascicles but also in regions where partial volume effect between WM and isotropic tissue occurs. In this work, we proposed a modified damped version of the Richardson Lucy (RL) spherical deconvolution approach to reduce instabilities and spurious components also in these regions.

**Theory:** The presence of isotropic compartments introduces a non-zero background that partially deactivates any non-negativity constraint and makes deconvolution process more susceptible to ringing effects and noise instabilities. Fig. 1 shows 2D profiles of a simulated fiber crossing system in presence and absence of a isotropic compartment (D=0.8 $10^{-3}$ $mm^2/s$, volume fraction = 0.5). Even assuming the exact fiber response and a noise-free case, when the isotropy is present, the solution, although non-negative, is free to oscillate below background level (dotted line). On the contrary, the solution obtained without the isotropic contribution is correctly recovered as non-negative constraint is sufficient to prevent any oscillation. In order to reduce these instabilities we applied a modified version of the "damped" RL algorithm as proposed by White for the original Poissonian RL algorithm in astronomic images [6]. Here, on each voxel, damping is applied taking into account the dynamic range of the recovered fiber orientations [7]. On high amplitude fiber components, a weak or no damping is applied preserving angular resolution. On low amplitude fiber components, a stronger damping is applied, reducing more probable spurious spikes and preventing noise amplification effects as iterations proceed. The proposed algorithm in matrix-vector notation is

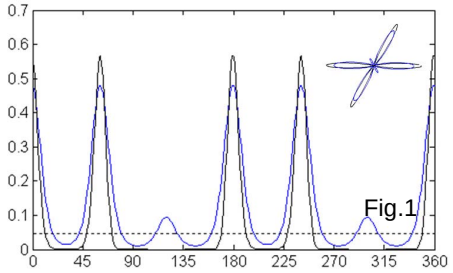

Fig.1

$$\mathbf{f}^{(k+1)} = \mathbf{f}^{(k)}\left(1-\mathbf{u}\left(\frac{\mathbf{H}^T\mathbf{s}-\mathbf{H}^T\mathbf{H}\mathbf{f}^{(k)}}{\mathbf{H}^T\mathbf{H}\mathbf{f}^{(k)}}\right)\right) \quad \text{with} \quad \mathbf{u}=1-\lambda\mathbf{r} \quad \text{and} \quad \mathbf{r}=\left(1-\frac{(\mathbf{f}^{(k)})^{\nu}}{(\mathbf{f}^{(k)})^{\nu}+\eta^{\nu}}\right)$$

where **f** is the estimated fiber orientation (nx1) vector, **s** is the HARDI sampled data (mx1) vector, **H** is the corresponding (mxn) circulant matrix [4]. **u** is a vector (nx1) that performs a damping on each **f** element. For $\mathbf{u}\to 1$ the algorithm is the same as in [4]; while for $\mathbf{u}\to 0$ the algorithm is reduced to $\mathbf{f}^{(k+1)}=\mathbf{f}^{(k)}$; $\eta$ and $\upsilon$ are two scalar damping parameters; $\eta$ is a threshold value acting on the amplitude of the recovered solution [7]. Moreover, to apply different levels of damping in different brain areas, a scalar term $\lambda$ is defined as $\lambda=1-4std(s)$, assuming a linear relation between anisotropy and standard deviation of the HARDI signal.

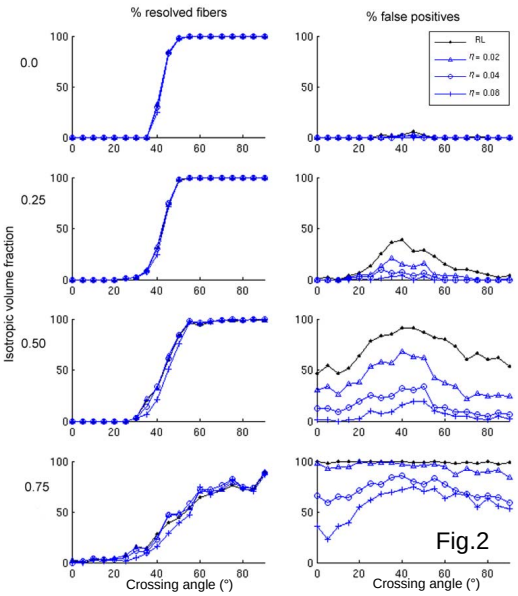

Fig.2

**Methods**

Simulations : Three-compartment configurations with different volume fractions were simulated: two identical anisotropic compartments (D=[1.7 0.2 0.2] $10^{-3}$ $mm^2/s$, $f_1=f_2$), crossing angles from 0° to 90°, one isotropic compartment (D=0.8 $10^{-3}$ $mm^2/s$, $f_3$=0, 0.25, 0.5, 0.75). $SNR_{b=0}$=20. 100 realizations were simulated for each configuration.

In vivo: A normal brain was acquired on a 3T Philips Intera scanner (Philips Medical System, Best, The Netherlands) with: TE/TR=74/14000 ms, FOV=240x240 mm, slices=40, slice thickness=2.5 mm, matrix=128x128, NEX=1, SENSE factor=2, bvalue=3000 s/$mm^2$, # DWdirections=60; Scan time=16min.

Deconvolution has been performed both with the standard RL approach and with the proposed method, imposing as fiber response a tensor equal to [1.5 0.3 0.3] $10^{-3}$ $mm^2/s$ and 200 algorithm iterations [4]. Different choices of $\eta$ have been applied, $\upsilon$ was set equal to 8.

**Results and Discussion** As shown in fig. 2, the number of resolved fibers (two distinct local maxima inside two 20° cone along fiber directions) is not affected by the damping procedure showing a similar angular resolution in both algorithms. Only with a threshold of $\eta$=0.08 the number of resolved fibers starts to decrease. On the other hand, the number of realizations with false positives (local maxima not corresponding to real orientations) is clearly reduced using the damped algorithm. In fact, with an isotropic volume fraction equal to 0.5, in the worst configuration, with the standard RL algorithm the percentage of false positives was 91%, while in the damped algorithm, imposing a threshold value of $\eta$=0.04, the false positives were only 34%. In fig 3 results from the in-vivo dataset are shown. White matter organization is clearly visible, showing several crossing fiber voxels coherently with anatomy and without evident spurious components. Moreover, in isotropic tissues the algorithm showed smoothed profiles correctly suggesting the absence of any fiber orientations.

In conclusion, the proposed algorithm is able to reduce the number of spurious components in regions with partial volume effect between WM and isotropic tissues without affecting angular resolution of the main fiber orientations.

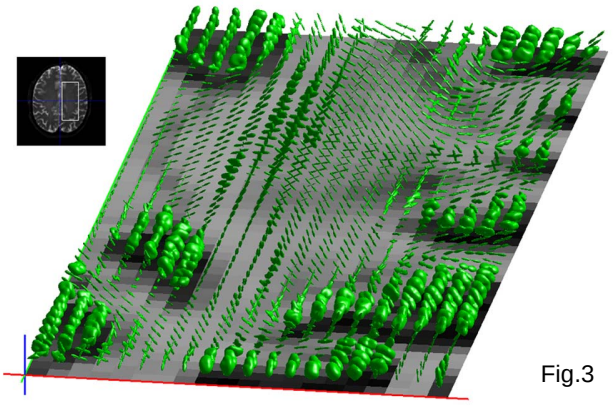
Fig.3

**References** [1] Tournier JD *et al*. NeuroImage 23:1176-1185 (2004); [2] Anderson AW, MRM 54:1194-1206, (2005) [3] Alexander DC Proc. IPMI (2005); [4] Dell'Acqua F *et al.* IEEE TBME 54(3):462-472, (2007); [5] Tournier JD *et al*. NeuroImage 35(4):1459-72, (2007); [6] RL White Proc. STScI : 104-110 (1993); [7] Dell'Acqua F *et al.* Proc. 15th ISMRM: 1473 (2007);