In vivo cardiac NMR Diffusion Weighted Imaging(DWI) for the human heart: improved quantification of FA and MD by edge-preserving regularization

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Purpose:

The purpose of this study is to apply an edge-preserving regularization technique to Cardiac Diffusion Weighted Imaging (DWI) data. Our aim is to improve in vivo DWI parametric maps quality towards clinical purposes.

Introduction:

Diffusion weighted imaging in the heart is greatly affected by contractile motion and remains challenging to date [1]. Accurate diffusion measurements require high diffusion encoding gradients and longer echo time that decrease signal strength and thus reduce significantly image quality. This results in diffusion weighted (DW) images corrupted by high level noise, which propagates to parameters computed from the diffusion tensor (e.g. fractional anisotropy, mean diffusivity). To overcome this drawback, we propose to perform edge-preserving regularization on the DW images. This is an effective intensity smoothing scheme which has the following characteristics: (1) image boundaries become sharper; (2) intra-region smoothing occurs preferentially over inter-region smoothing; (3) no spurious details is produced in the images.

Theory:

Given a DW image $y \in \Re^N$ corrupted by noise (N denotes the number of pixels) the goal is to find the noiseless DW image x^0 according to the model $y = x^0 + \eta$, where η is assumed to be white Gaussian noise. The solution to this problem is defined as the minimum of a cost functional $U : \Re^N \to \Re$ of the form $U(x) = \|y - x\|^2 + \lambda\Phi(x)$, where the prior term Φ favors the formation of piecewise smooth configurations and the smoothing parameter λ balances the effect of Φ with the data fidelity term $\|y - x\|^2$ [2].We adopt the widespread prior model $\Phi(x) = \sum_{c \in V} \phi(D_c(x))$,where V denotes the pixel lattice supporting the image and D is a linear approximation of the first-order derivative. The function ϕ is defined by $\phi(t) = 2\log(\cosh(t))$ to ensure the convexity of U while reducing the smoothing effect in the vicinity of discontinuities. The optimization problem involved in minimizing U is tackled with the deterministic half-quadratic regularization algorithm proposed in [3]. In [4], we prove that our regularization method provide equivalent results to the most recent state of the art approaches (e.g. [5], [6]).

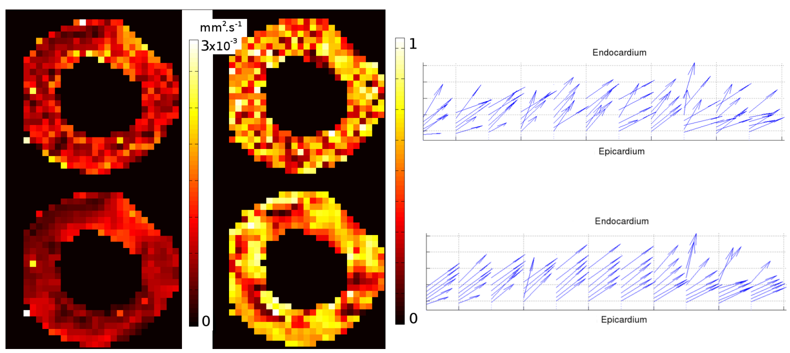

*Figure 1: Impact of the regularization on in-vivo data. **Left to right**: parametric maps ordered as follows: MD, FA and PDD field. **Top**: parameters computed from the tMIP raw DW images. **Bottom**: parameters computed from the tMIP regularized DW images*

Method:

Ex-vivo and in-vivo experiments were conducted with our Spin Echo time-Maximum Intensity Projection (tMIP)-DWI method, on a 1.5 T clinical scanner (Siemens, Avanto). Its principle is to remove contractile-motion related signal-loss by applying temporal maximum-intensity projection on a series of single-shot images over a time window that covers end diastole (resolution: 2.2x2.2x3.2mm3, matrix FOV: 281.6 x 70.4 mm², DW parameters b: 50 and 200s/mm²).

To validate the overall process of tMIP-DWI acquisition with edge-preserving regularization, we first compared on ex-vivo human hearts tensor results obtained from our method and from the well established DWI Stejskal-Tanner sequence[7] (experiments in agreement with forensic institute ethic committee). We used appropriate ex-vivo values b=0 and b=1000s/mm² on both sequences, with 6 diffusion encoding directions for our sequence and 12 directions for the Stejskal-Tanner. We averaged both acquisitions 4 times and used the same algorithm to compute the diffusion tensor and its associated parameters. Then, we applied the method to in-vivo experiments, conducted on ten healthy volunteers and ten patients, to obtain a tensor image in early diastole within less than 4min. Subjects were allowed to breath freely for a maximum of comfort. Images were acquired during expiration. They were segmented afterward with a semi-automatic method based on level-set contour propagation [8] and then registered with a non rigid b-spline algorithm [9].

The effect of regularization is studied at different steps of the DT-MRI processing pipeline: parametric maps (fractional anisotropy, mean diffusivity) and principal diffusion direction field. It enables us to see how well the regularization method inhibits noise propagation along the DT-MRI processing pipeline. Fractional anisotropy (FA[10]) measures deviation from isotropy and reflects the degree of alignment of cellular structures within fiber tracts ; mean diffusivity (MD[10]) measures average molecular motion and principal diffusion direction (PDD) field indicates how fibers are organized locally (at each pixel).

Results and Discussion:

Figure 1 compares the parameters computed from the tMIP DW images with the parameters computed from the regularized tMIP DW images for in-vivo experiments. Edge-preserving regularization allows maintaining the contrast of the MD map while substantially decreasing noise on the FA map. Also, the PDD field appears more coherent when computed from regularized data: the fiber orientation variation through the left ventricular wall is emphasized. To further validate, we consider ex-vivo experiments (see Fig 2): 3-D edge preserving regularization applied to tMIP DW images enables to provide equivalent results to Stejskal-Tanner DW images. Note however, that if Stejskal-Tanner sequence displays less distortion, it does not apply to in-vivo human heart.

Postprocessing of DWI data using 3-D edge-preserving regularization stands as an efficient solution to reduce noise due to in-vivo cardiac DWI constraints, therefore improving DWI data interpretation toward clinical purposes.

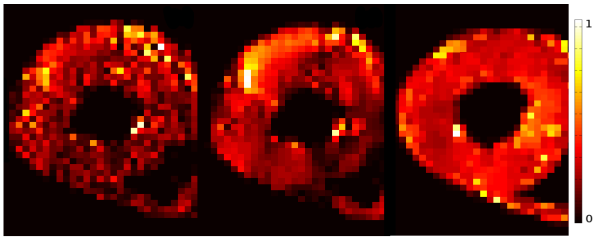

*Figure 2: Impact of the regularization on ex-vivo data. **Left to right:** FA computed from the raw tMIP DW images, the regularized tMIP DW images and from the raw Stejskal-Tanner DW images.*

References:

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