Cardiac Diffusion Tensor Imaging: Adaptive Anisotropic Gaussian Filtering to Reduce Acquisition Time

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Targeted Audience: Biomedical engineers, researchers and biomechanical engineers.

Purpose: Myocardial fiber structure exhibits a complex counter-directional helical geometry, characterized by its helical angles (HA)[1]. HA is estimated as the angle between the circumferential axis and the projection of the fiber onto the circumferential longitudinal plane. These angles smoothly change from a negative to a positive helix from the epicardium to the endocardium. The counter-directional helical structure is energetically efficient and plays an important role in understanding the electrical and mechanical properties of the heart[2]. Furthermore, pathophysiological information of different cardiac disease states can be characterized by HA[3]. Thus, there exists a need to non-invasively quantify HA which can be determined using diffusion tensor imaging (DTI)[4]. The evaluation of HA measured from DTI depends on the accuracy of the diffusion tensors which in turn depend on the number of diffusion-encoding directions (DED) and the signal to noise ratio (SNR), which can be increased by increasing the total number of excitations (NEX). However, increasing SNR and DED also increases the total acquisition time (TA). The purpose of this study is to reduce TA by implementing a three dimensional (3D) adaptive anisotropic Gaussian filter (AAGF) on the diffusion tensors to accomplish the same robust HA measurements as achieved using larger NEX and DED.

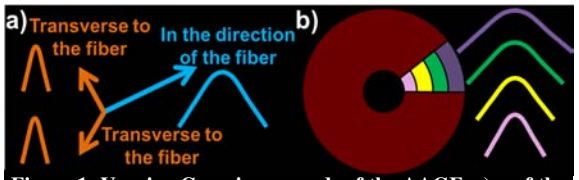

Figure 1: Varying Gaussian spreads of the AAGF. a) σ of the TF larger in the direction of anisotropy. b) σ of the TF increases with increasing radius.

Methods: ***Filter Design:*** The 3D AAGF has been designed such that its transfer function (TF) is rotated in the direction of the fiber with the spread (σ) of the Gaussian function being maximum in this direction (Figure 1(a)). The TF is also designed to be radially dependent, i.e., σ in the direction of the fiber increases as a function of the radial distance (Figure 1(b)). This renders the filter to be both anisotropic and adaptive in nature. ***Acquisition:*** Ex-vivo DTI was performed on a formalin fixed porcine heart in a 3T MRI scanner (Tim Trio, Siemens Healthcare). A two dimensional bipolar diffusion-weighted spin-echo based echo planar imaging sequence was applied to acquire multi-slice short axis views of the heart. Thirty three scans were performed with different combinations of NEX and DED. Imaging parameters included: DED [NEX]:= 12 [1-20], 30 [1-8], 64[1-4], 256 [1]; TE/TR=90/7000ms; slice thickness=2mm; matrix=128x128; FOV=256x256mm^2; b-values=0,1000s/mm^2; slices=42. ***Analysis:*** The acquired images were masked to segment the left ventricle (LV). Custom-built software written in Matlab (Mathworks, Natick, MA) was used to obtain the diffusion tensors, implement the AAGF and finally estimate HA based on both the filtered and unfiltered tensors. From our initial experience, a 11x11x11 3D window was chosen for the filter and σ along the radial distance was incremented with a step size of 0.5. Line profiles (LP) were computed both on the filtered and unfiltered data, along 16 equally spaced (11.25^o apart) transmural regions (Figure 3) on the free wall of the LV to evaluate the effect of filtering on HA.

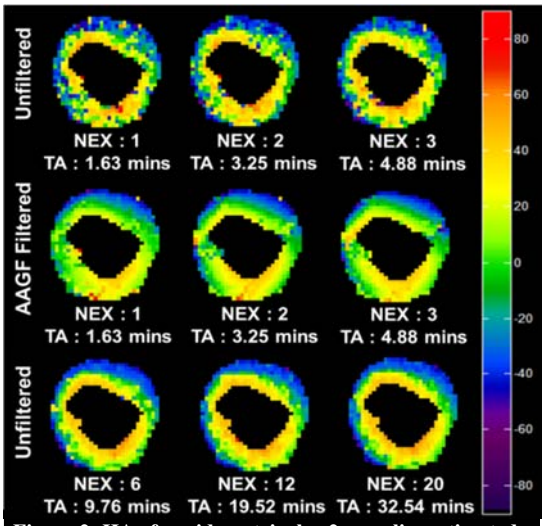

Figure 2: HA of a mid-ventricular 2 mm slice estimated from a 12 DED DTI acquisition.

Results: Figure 2 shows HA from a mid-ventricular slice in the LV obtained from a 12 DED acquisition. The 1st row shows HA estimated from unfiltered diffusion tensors. The 2nd row shows their filtered counterparts. The 3rd row illustrates HA with increasing NEX. We have observed that the AAGF filtered results in the 2nd row (lower SNR, lower TA) are nearly consistent with the unfiltered HA in the 3rd row (higher SNR, higher TA). The 3rd row also demonstrates that increasing NEX above a certain value (6 in case of 12 DED acquisition), just increases the TA but does not provide any noticeable improvement in the HA transition in the transmural direction. Figure 3 shows three different LP along the free wall of the LV which indicates that the AAGF filtered HA LP (solid line) have a much smoother transition from the epicardium to the endocardium compared to the unfiltered HA LP (dashed line). A similar trend was observed in the entire LP's in the other slices and also in the other datasets acquired with different DED (not shown here). Furthermore, these HA are in agreement with previous published results[4]. The variation of HA from epicardium to endocardium along the transmural distance is also consistent with previous ex-vivo histopathology results[2].

Conclusion: Our results demonstrate that by applying a 3D AAGF we can robustly estimate HA with almost a 60% reduction in scan time. However, further investigation is necessary to optimize our filter and establish our technique.

References:

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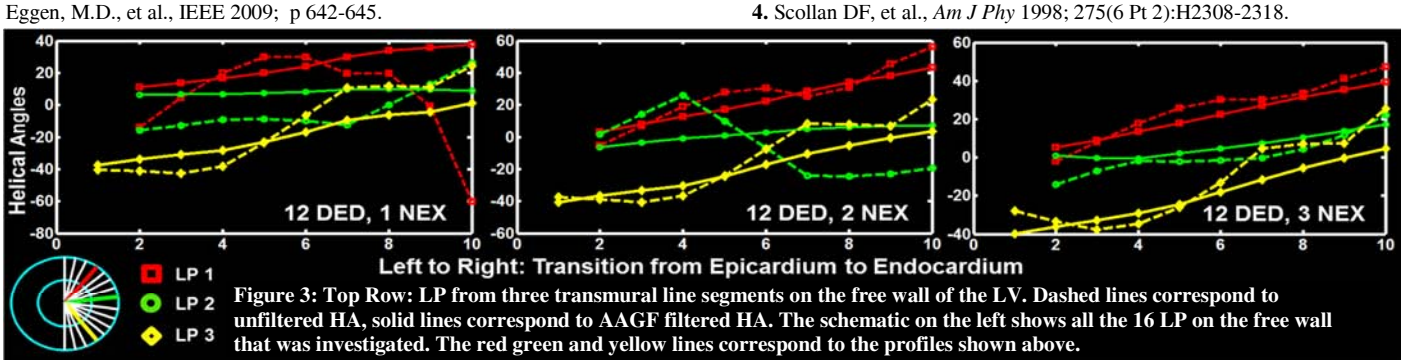

Figure 3: Top Row: LP from three transmural line segments on the free wall of the LV. Dashed lines correspond to unfiltered HA, solid lines correspond to AAGF filtered HA. The schematic on the left shows all the 16 LP on the free wall that was investigated. The red green and yellow lines correspond to the profiles shown above.