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CARDIAC MRI DERIVED EPICARDIAL FAT MAPS TO ASSIST VT ABLATION PROCEDURES FOR SUBJECTS WITH IMPLANTABLE DEVICES

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ABSTRACT

Cardiac electroanatomical mapping (EAM) to locate myocardial scar substrate during ventricular tachycardia (VT) ablation therapy is currently faced with a major limitation. The presence of epicardial fat leads to false-positive low-voltage maps that may be interpreted as myocardial scar. Pre-procedural cardiac magnetic resonance (CMR) imaging can produce images of the heart in which both cardiac geometry and epicardial fat are particularly conspicuous. The goal of the present work is to build CMR based geometric models composed of a shell of the cardiac anatomy and epicardial fat. Herein, we first defined a black-blood imaging protocol that is feasible in patients with implanted devices. Then a semi-automated image segmentation method was developed and applied to extract ventricular surface anatomy and epicardial fat maps from which three-dimensional surfaces were built. The results suggest that deriving maps of epicardial fat from MRI data is feasible, accurate when compared to expert observers, and suitable for integration in a clinical catheter-based ablation procedure.

Index Terms— Cardiac MRI, Implantable Cardioverter Defibrillators, Image Segmentation, Epicardial Fat, Image Based Modeling

1. INTRODUCTION

Myocardial infarcts (MI) are a leading cause of worldwide morbidity and mortality. The resulting ventricular scars form the substrate that is typically responsible for ventricular tachycardia (VT) and subsequently lethal ventricular fibrillation (VF). To limit the recurrence of VF, VT ablation is a minimally invasive catheter procedure that delivers RF energy within and around the myocardial scar and targets mitigating VT [1]. Ablation procedures are supported by means of epicardial and endocardial electroanatomical mapping (EAM). EAM produces three-dimensional surface voltage and activation time maps and provides significant assistance when determining ablation sites [2, 3, 4]. In particular, scar tissue correlates to areas of low voltage and thus defines a possible ablation site which is then further evaluated as a target

for ablation. However, one major limitation of EAM is that the presence of an epicardial adipose tissue (EAT) leads to similarly low voltages as scar tissue. EAT thus creates false positives and subsequently may lead to unnecessary ablation and a prolonged procedure.

Cardiac MRI is a useful pre-procedural examination for this patient population and due to its flexible image contrast it produces a variety of images for assessing cardiac anatomy and function. Late gadolinium contrast enhanced MRI (LGE-MRI) presents the method of choice for identifying myocardial scar morphology. In addition, MRI is able to deliver images in which both ventricular geometry and adjacent EAT are especially conspicuous. However, one challenge is the presence of an implantable pacemaker or cardioverter defibrillator (ICD), which is conventionally contraindicated for an MRI exam. Nevertheless, experienced clinical centers routinely scan these patients under close supervision and with specialized imaging protocols [5, 6, 7]. Furthermore, recent breakthroughs in reducing ICD related image artifacts consolidate the application of MRI for VT patients [8, 9]. Integrating image based three-dimensional maps of ventricular anatomy and EAT into the VT mapping procedure may compensate for this limitation.

Dickfeld *et al.* [10] investigated the integration of contrast enhanced MRI derived scar maps into the VT ablation procedure for patients with ICDs. The fusion of MRI and EAM data reduced the number of false positives (low voltage recordings) in $4.1 \pm 1.9\%$ of the mapping points with a surface voltage < 1.5 mV. These findings suggest that MRI can be safely performed in patients with ICDs and the image output can be used to build 3D scar maps, which facilitate VT ablation by providing additional anatomical guidance. Similarly, Roujol *et al.* [11] introduced advanced methods for registering MRI derived scar information and EAM voltage maps. Approaches for the integration of information about cardiac epicardial fat into the VT ablation procedure are more sparse. Though, Nelson *et al.* [12] have shown that cardiac adiposity can accurately be assessed by means of clinical cardiac MRI sequences. In contrast to including MRI information, van Huls van Taxis *et al.* [13] carried out a patient study (N=28), where pre-procedural multi-detector computed tomography (MDCT) images were used to model epicardial fat and coro-

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nary arteries. The models were fused with the EAM output post-procedurally to assess correlation of voltage mapping with epicardial fat thickness. The results indicated that areas of 7 mm fat thickness were associated with ablation failure, due to false positive low voltage mapping caused by the interfering epicardial fat layer.

Despite the efforts presented above, no approach for integrating cardiac MRI derived EAT maps has been reported. Therefore, the overall objective of this work was to build CMR-based geometric models of ventricular anatomy and epicardial fat, which can be integrated into the VT ablation procedure. To account for the commonly present cardiac pacemaker or ICD, we subdivided the project into two major parts: (1) to define and evaluate a cardiac MRI protocol for patients with an ICD that provides complete cardiac coverage with excellent EAT-myocardium-blood contrast; (2) to develop and implement an image processing pipeline that creates MRI-based three-dimensional geometric models of the ventricular epicardium and EAT.

2. METHODS

2.1. Image Acquisition

Subsequent to obtaining informed consent healthy volunteers (N=10) underwent an IRB approved cardiac MRI exam to acquire a stack of 21-24 short-axis black blood images without inter-slice gaps covering both ventricles. All scans were performed on a 1.5 T MRI scanner (Siemens Avanto). Typical parameters for the black blood turbo spin echo (BB TSE) sequence were as follows: TR: 1192 ms, TE: 49 ms, ETL: 40, flip angle: 150°, image matrix: 256 x 200, in-plane resolution: 1.4 x 1.4 mm, and slice thickness 6 mm. 2D image acquisition was performed during breath-holding at end-expiration and at end-diastole. The respiratory pattern (in non breath-hold phases) was monitored using a pencil beam navigator. Depending on the patient's breath-hold ability, acqui-

sition of multiple slices within one breath-hold is possible. Typically, a feasible breath-hold duration (15 s) allows for the acquisition of three image slices. Therefore, the likelihood of inter-slice misalignment is reduced compared to single slice per breath-hold protocols. The duration of complete image stack acquisition averaged 2.5 min.

To evaluate the impact of an ICD on image quality, an ICD was attached to the anterior MRI receiver coil. To ensure appropriate external ICD placement comparable to device location in a patient, we acquired a single HLA slice using an inversion recovery (IR) pulse sequence that is conventionally used for late gadolinium enhanced (LGE)-MRI scar imaging. The extent of ICD induced high intensity image artifacts were qualitatively compared to typical device artifacts in patient images and the device was repositioned as needed.

2.2. Image Analysis

Based on a short axis stack of 2D BB TSE images, regions of left ventricular blood cavity (LV ENDO), myocardium (LV MYO), and right ventricle (RV) were segmented using a semi-automated multi-label random walk approach [14]. First, BB TSE image data were preprocessed by means of automated window leveling (threshold intensities below the 10th and above the 90th percentile) and anisotropic diffusion processing to increase image homogeneity, while preserving valuable edge information. Second, seeds labeling each region to be segmented were selected using a interactive graphical user interface. Third, the random walk output was computed and multi label 2D masks were post-processed through morphological image closing to smooth boundaries. The overall bi-ventricular boundary composing both the LV and RV was obtained by fusing LV ENDO, LV MYO and RV masks.

EAT segmentation based on the same 2D BB TSE image data poses several image analysis challenges given that: (1) Fat may or may not exist and the fat distribution on the epicardium does not follow a particular shape or pattern, which limits the (spatial) assumptions one can incorporate when designing the processing pipeline; and (2) Epicardial fat is sometimes indistinguishable from pericardial fat because image resolution limits conspicuity for the pericardial sac.

The epicardial fat segmentation in our framework implements a spatially constrained region-growing method that uses the segmented epicardial border in combination with manually placed seeds for the initialization purposes. The algorithm can be summarized as follows: (1) Define a ROI by dilating the epicardial boundary assuming the EAT thickness does not exceed 10mm, which is constrained by the structuring element used for morphological dilation; (2) Select EAT seeds within a ROI via an interactive user interface; (3) Region grow from each seed point while constraining growth to the defined ROI (homogeneity criteria set to threshold $T_h = 0.1$, after intensity normalization to $[0, 1]$ scale).

Fig. 1: Processing pipeline

Subsequent to the segmentation of epicardium and EAT with binary masks as output, three-dimensional surface meshing was performed. Epicardial boundaries were triangulated by extracting boundary points as image pixel coordinates from the set of 2D masks. Upsampling the boundary points through linear interpolation results in an equivalent number of points within each image slice of the three-dimensional point cloud. Point coordinates were then scaled to account for in-plane resolution and slice thickness. Finally, iterative triangulation found corresponding faces that were defined by the vertex indices (point cloud). In contrast, the fat segmentation output was modeled using existing Matlab methods to define a three-dimensional isosurface.

2.3. Validation of EAT segmentation

The output of EAT segmentation using the proposed semi-automated algorithm was validated through comparison with the manual tracing (MT) output performed by an independent expert. To assess inter-observer variation, two observers used the semi-automated tool (U1 and U2). Three Dice similarity indices are computed for each image slice: $D(MT, U1)$, $D(MT, U2)$ and $D(U1, U2)$.

3. RESULTS

	$D(MT, U1)$	$D(MT, U2)$	$D(U1, U2)$
1	0.66 (0.04)	0.66 (0.06)	0.91 (0.09)
2	0.60 (0.10)	0.61 (0.08)	0.76 (0.20)
3	0.42 (0.15)	0.35 (0.12)	0.56 (0.27)
4	0.57 (0.09)	0.58 (0.11)	0.63 (0.21)
5	0.48 (0.21)	0.49 (0.19)	0.63 (0.16)
All	0.55 (0.15)	0.53 (0.16)	0.69 (0.22)

Table 1: EAT segmentation validation: Average Dice indices and standard deviation for each study (1-5), and overall (All).

Exemplary results of 2D BB TSE image data as well as epicardial and EAT segmentation are shown in figure 2. External ICD placement was feasible in all volunteer scans and the IR control images demonstrated high intensity image artifacts similar to those typically observed in patients with ICDs. Images displayed ICD related void artifacts, which do not interfere with the cardiac region of interest. Excellent blood pool to myocardium contrast in basal and mid-ventricular image slices is presented. Contrast and sharp region boundaries worsened in the apical image slices due to well-known slow flow image artifacts. Moreover, the image intensity of EAT surrounding the heart clearly differs from the blood pool and myocardium pixel intensity, which facilitates image segmentation. Random walk segmentation results (column a) of three cardiac regions defined the a-priori information for segmenting EAT. Semi-automated segmentation of EAT (column b,c) output thin layers of fat attached to

Fig. 3: Correlation of Dice index and the number of (manually traced) EAT pixels per slice. Colors represent each volunteer study.

Fig. 4: Rendering of reconstructed surface meshes composing the epicardial shell (blue mesh) and overlaying EAT (yellow).

the epicardial boundary. For comparison, the corresponding manual segmentation results were plotted in figure 2 column d).

To demonstrate the semi-automated segmentation performance, Dice similarity coefficients are given in table 1. Figure 3 shows the correlation between the segmented fat area and computed Dice value per slice. Figure 4 show reconstructed surface meshes composing both epicardial shell and EAT of two different volunteers.

4. DISCUSSION

The 2D BB TSE protocol produced images suitable for defining both cardiac shell anatomy and EAT. The short acquisition time and feasible breath-hold duration allows for integration into a standard clinical MRI protocol for patients with ICDs. However, visual separation of EAT and peripheral fat remains challenging. This directly impacts the ability to segment EAT both manually and semi-automatically. Dice similarity increased with EAT area; the smaller the EAT layer, the easier it was to slightly misplace the initialization seeds or to manually delineate. A comparison of the segmented areas (semi-automated vs. manual) in figure 2 demonstrated high simi-

Fig. 2: Two BB TSE short axis slices in top and bottom row, respectively. Segmentation of cardiac boundaries using random walk (col a), segmentation of EAT using semi-automated algorithm (col b,c), and manually traced EAT (col d).

larity of the overall EAT location and shape, but differences in EAT thickness lead to moderate Dice coefficients. Inter-observer Dice coefficients were higher, indicating the tool is robust.

The task of accurate EAT segmentation remains challenging and simply showing the presence of EAT within each anatomical segment, rather than its thickness, may work as an alternative. In contrast, if segmentation accuracy can be improved quantitative EAT visualization through color coded fat thickness maps may be more well suited to the clinical application.

5. CONCLUSION

This work presents an image acquisition and processing pipeline for building three-dimensional maps of epicardial surface geometry and adjacent fat pad maps derived from cardiac MRI short axis black blood image data. Moving forward, the framework will be augmented to include myocardial scar information. The segmented results then need to be evaluated during clinical procedures wherein EAM voltage recordings can be compared to the underlying tissue type.

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