CONICAL DEFORMABLE MODEL FOR MYOCARDIAL SEGMENTATION IN LATE-ENHANCED MRI

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ABSTRACT

This paper presents a conical 3D deformable template for fully automatic and robust segmentation of late-enhanced MRI (LE-MRI) datasets. The proposed technique has several advantages over existing works. Firstly, it uses a thick-walled conical model that is suitable to derive fully automatic and reliable initialization by taking into account potential short-axis misalignments. Furthermore, it uses to its advantage the geometrical and appearance properties of the blood pool to decouple the endocardial and epicardial segmentations. The final epicardial result is obtained using thickness smoothness measures constrained on the initial robust segmentation of the endocardium. Detailed validation using 20 LE-MRI datasets and comparison to previous work demonstrates that the technique is reliable and promising for clinical assessment of LE-MRI data.

Index Terms— Myocardial segmentation, LE-MRI, 3D model, deformable model, slice alignment correction

1. INTRODUCTION

Correct assessment of the extent of viable and nonviable myocardial tissue is necessary for the diagnosis and treatment of cardiovascular diseases [1]. To this end, Late-Enhanced MRI (LE-MRI) has established itself as a standard imaging protocol in clinical practice to localize and quantify myocardial scar tissue [2, 3].

Viability assessment can be divided into two steps: segmentation of the myocardium (endocardium and epicardium contours) on every slice of the LE-MRI short-axis stack, and posterior quantification of tissue viability. Due to the specificity of the image appearance in LE-MRI (see Fig. 1) , with the presence of inhomogeneities because of the abnormalities and scar tissues, the segmentation step is still performed manually in clinical practice. However, manual delineation is time consuming and varies significantly between observers. To rapidly and consistently assess the increasingly large MRI data available in clinical practice, as well as to eliminate errors and inconsistencies due to user intervention, fully automatic extraction of the myocardium with the incorporation of automatic initialization is critical.

To achieve this, two main strategies can be adopted. Firstly, the myocardium can be segmented on the cine-MR images followed by the application of a suitable registration stage onto the LE-MRI dataset. Several difficulties arise with this approach: due to inconsistencies in patient position and breath-hold during scanning, significant short-axis misalignment often arise between the cine- and LE-MRI datasets. This issue is currently not addressed in the literature and it constitutes a main motivation of the work proposed in this paper. Other spatial and temporal variations can also appear between the two protocols and this has motivated the use of additional non-rigid registrations such as in [5] and [6] to cope for the associated inaccuracies. But these techniques depend on the quality of the registration, which can suffer from the differences in image appearance and the presence of artifacts due to the scar tissues in LE-MRI. A second strategy is to directly segment the LE-MRI datasets which would resolve the issues described above, in particular the spatial and temporal discrepancies that arise between the cine- and LE-MR scans. However, the task is challenging due to the inconsistencies in geometry and image quality. In particu-

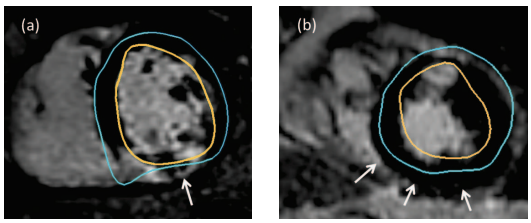


Fig. 1. Results applying an ASM method [4] in LE-MRI (a) Scar effect (b) Slice shift effect

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lar, the appearance of the myocardium is complicated by the presence of scar tissue, which can affect the accuracy and robustness of automatic techniques. Fig. 1 provides two examples of the application of the well-known active shape models (ASM) [4]. Despite its use of prior knowledge, it is affected by the presence of scar inhomogeneities (Fig. 1(a)) and shift due to slice misalignment (Fig. 1(b)), shown by the arrows.

Due to these difficulties, very little has been published on fully automatic segmentation of the myocardial borders in LE-MRI. Amongst more recent attempts, Ciofolo et al. [7] use a geometrical template to find the myocardial contours by combining basic shape, region and gradient information. While the idea is sound, the method is implemented in 2D. Furthermore, the method requires not less than six free parameters to tune, which makes it difficult to identify the optimal combination for the segmentation task, and which makes the method time consuming in practice.

Based on these observations, we propose an alternative method that can fully automatically and robustly segment the myocardium in LE-MRI. The method uses a three dimensional reliable initialization scheme based on a thick-walled conical template that can cope furthermore with short-axis misalignments. In addition to this, it uses to its advantage the geometrical and appearance properties of the myocardium in LE-MRI, by decoupling the endocardial and epicardial searches. Detailed validation of the method is carried out with experiments based on 20 LE-MRI datasets.

2. METHOD

The proposed method is based on the use of a thick-walled conical model to represent the left ventricular morphology and to guide the image search procedure (Fig. 2(a)). As detailed in subsequent sections, this template has several advantages for myocardial segmentation in LE-MRI, in particular its ability to deal with inconsistencies in the left-ventricular axis and its suitability for constraining the search of the endocardial and epicardial boundaries. In the proposed method, and unlike existing methods that use a single image search procedure, we decouple the segmentation of the endocardium and epicardium with the aim at using to our advantage the geometrical and appearance properties of the blood pool and of the myocardium. With this strategy, 3D endocardial initialization (Sec. 2.1) and segmentation (Sec. 2.2) is achieved by taking into account the stronger contrast of the blood pool and by avoiding the inconsistencies in image appearance due to scar tissue. This is then followed by the segmentation of the epicardium (Sec. 2.3) constrained by the more robust previous localization of the endocardium, with the incorporation of smoothness measures with respect to the thickness of the myocardium.

2.1. 3D endocardial initialization

Automatic initialization in LE-MRI is challenging because of variability in the left-ventricular axis, which is both subject-specific and due to short-axis misalignments. To solve this, and since information regarding ventricular axis is irrelevant to scar quantification, we use a conical template (see Fig. 2(a)) that allows to realign all short-axis slices. Moreover, it is very critical that the derived initialization is reliable independently of the scar tissue appearance. Therefore, we use the endocardium for initialization as the blood pool has stronger contrast, and scar tissue will not interfere at this stage.

The appearance of the myocardium (dark and circular) makes it detectable using a Laplacian of Gaussian template ($\mathbf{T}$) modeling a dark ring, similarly to [7] (see Fig. 2(b)), by convolving it with each slice:

$$[\mathbf{z}_i, r_i, w_i] = \arg\max(\mathbf{T}(\mathbf{z}, r, w) * \mathbf{I}) \quad (1)$$

where r and w are the radius and thickness of the template respectively, and $\mathbf{z}$ is the location of the center of the template.

A set of radius and width values are tested and the combination that provides the best grey-level matching is extracted (see Fig. 2(c)). The set is changed according to the position of the slice that is being segmented (given by the DICOM header). We have also preprocessed the images by using a sigmoid function to increase the contrast of the blood pool during endocardial initialization.

The 3D volumetric information is used in the proposed technique to detect and correct potentially erroneous initializations. More specifically, after a first round of initialization, the median value of the obtained centers for all the slices of the volume is computed. A second round of initialization is then performed which is focused on a window around the obtained mean center and with an estimation of the possible radius. As shows in Fig. 3, erroneous slice initializations are obtained at the first iteration but they are successfully detected and corrected at next iteration of the initialization algorithm. Once a first estimation of myocardial position is found, the centroids are recomputed and the slices are aligned to match the 3D conical model (see Fig. 2(a)). This is a central feature of the proposed technique, since it reliably eliminates inaccurate initializations and corrects slice misalignment.

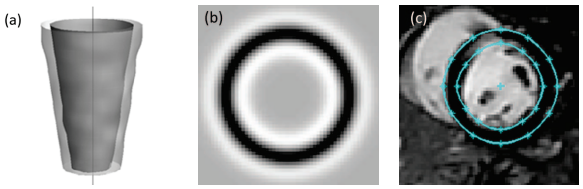


Fig. 2. (a) 3D Conical model (b) Laplacian of Gaussian template (c) Initial position, radius and width of the model

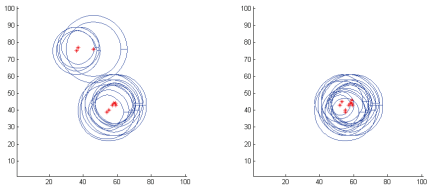


Fig. 3. Illustration of the two rounds of the initialization process for a patient. Each circle (blue) and center (red star) corresponds to a slice of a single volume. First round of initialization (left) and final initialization (right)

2.2. 3D endocardial segmentation

Instead of segmenting in one optimization both the endocardium and epicardium as achieved for example in [7], we propose to decouple the segmentation of the two boundaries to achieve the following advantages. Firstly, due to the contrast in the blood pool, we can reliably achieve a more robust segmentation of the endocardium independently of the inconsistencies in scar appearance within the myocardium. Subsequently, the endocardium will be used to constraint the subsequent epicardial segmentation by incorporating smoothness measures related to the myocardial thickness, with the goal to achieve robustness to the scar tissue appearance.

For the endocardial segmentation, the deformable conical template moves through the image space to minimize a specified optimization function. The first term in Eq.(2) eliminates solutions with inconsistent curvature information along the boundary, while the second term moves the contour towards the image features of interest, in this case by using gradient measures. The total objective function to minimize is:

$$E(\mathbf{I},\mathbf{l}) = \alpha \int_0^1 (|\mathbf{l}(s)''|^2)ds - (1-\alpha)\int_0^1 |\nabla \mathbf{I}(s)|^2 ds \quad (2)$$

where $\mathbf{l}$ is the endocardium line and $\nabla\mathbf{I}(s)$ is the magnitude of the image gradient along the endocardium line.

This equation ensures that the curvature of the myocardium has acceptable variability around the average curvature, to take into account the physiological properties of the left ventricular myocardium. Furthermore, it ensures that the contours are effectively attracted by the high contrast found in the blood pool.

After robust segmentation of the endocardium is achieved, a last slice realignment can be performed according to the conical model.

2.3. 3D epicardial segmentation

With the robust localization of the endocardium achieved in the previous segmentation, a good initialization of the epicardium segmentation is available and we can now deform the conical template by taking into account smoothness on the myocardial thickness as follows:

$$E(\mathbf{I},\mathbf{w}) = \beta \int_0^1 (|\mathbf{w}(s)''|^2)ds - (1-\beta)\int_0^1 |\nabla \mathbf{I}(s)|^2 ds \quad (3)$$

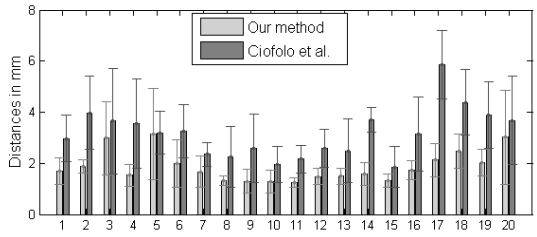
where $\mathbf{w}$ is the myocardial thickness and $\nabla\mathbf{I}(s)$ is the magnitude of the image gradient along the epicardium line.

It is important to note that each stage of the algorithm uses only one free parameter during the image search. This simplifies significantly the definition of the optimal value of the parameter.

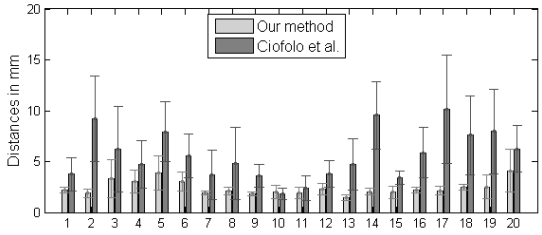
3. RESULTS

To assess the performance of the proposed method, a set of 20 LE-MRI datasets were acquired using a General Electric Signa CVi-HDx, 1.5T scanner (General Electric, Milwaukee, USA). The short-axis images were scanned with a LGE inversion recovery sequence (10 min after IV administration of 0.2 mmol/kg of Gadopentate Dimeglumine contrast). Relevant acquisition parameters included: 6 to 12 slices for each volume, a plane resolution of $1.56mm \times 1.56mm$, a slice thickness of 8 mm, and a spacing between slices of 8 to $10mm$.

The proposed algorithm was then applied to all datasets, along with the first step of our implementation of LE-MRI segmentation technique by Ciofolo et al. [7] for comparison. The results of the automatic approaches were compared with manual delineations of the datasets, which were performed by two separate observers. The accuracy of the extracted contours was evaluated by calculating the point-to-contour error and the Dice similarity coefficient (DSC) (measuring the degree of area overlap). These results are summarized in Table 1 for the initialization stage, the final segmentation and the



(a) Endocardial errors



(b) Epicardial errors

Fig. 4. Bar graph visualizing the segmentation errors (2D point- to-contour distance) for the endocardium and epicardium of 20 datasets with our method and our implementation of [7]

Table 1. Results automatic versus manual segmentations (*distances expressed in mm)

	Initialization			Segmentation			Ciofolo et al. [7]		
	ED Endo*	ED Epi*	DSC	ED Endo*	ED Epi*	DSC	ED Endo*	ED Epi*	DSC
Mean	2.24	2.75	0.68	1.86	2.41	0.78	3.18	5.64	0.66
Standard Deviation	0.80	0.75	0.08	0.60	0.70	0.07	0.96	1.62	0.07
Median	2.01	2.32	0.74	1.68	2.16	0.81	3.17	5.17	0.68
Maximum value	4.28	5.15	0.84	3.14	4.07	0.85	5.85	10.14	0.82
Minimum value	1.34	1.58	0.52	1.23	1.43	0.58	1.85	0.52	0.52

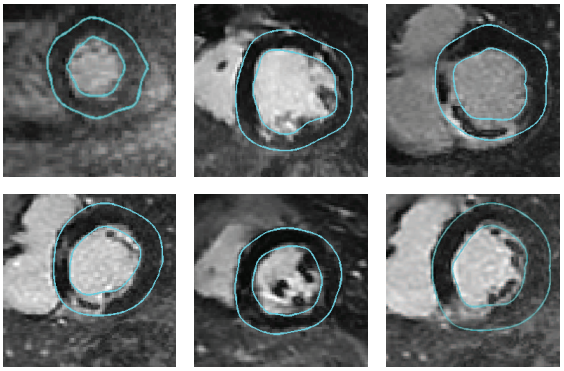


Fig. 5. Some segmentation results of our automatic method

existing method in [7]. Firstly, it can be observed that the proposed initialization is reliable throughout all datasets, with a maximal initialization error of $4.28mm$, which is the result of the introduced conical template that allows efficient use of the 3D image information during the procedure. Moreover, it is evident from Table 1 that the proposed segmentation is not only more accurate but also more consistent throughout all datasets, with clearly better average and standard deviation statistics. This performance is well illustrated in Fig. 4, where the individual endocardial (Fig. 4(a)) and epicardial (Fig. 4(b)) results are plotted for all datasets, showing significant improvement throughout all sample. This demonstrates the benefit of the proposed iterative initialization, as well as of decoupled approach that allows robust segmentation of the endocardium and effective subsequent constraining of the epicardial search. Some illustrations of the automatic segmentations are shown in Fig. 5, where it can be noticed how the obtained contours are resistant to the presence of scar tissue inhomogeneities. Finally, it is worth mentioning that the proposed technique is obtained on average 15 times faster than Ciofolo's method on the same machine. This is due to the decoupled constraints that require less parameters to tune and achieve improved myocardial localization.

4. DISCUSSION AND CONCLUSIONS

The key novelty of the proposed technique is the use of thicked-wall conical template for fully automatic initialization and segmentation of the myocardium in LE-MRI datasets. The three-step approach used in this work allows robust localization of the endocardium and effective constraining of the more challenging epicardial search. In addition, the method has only one parameter to optimize, which make it more robust to dataset specificities. The validation shows that the method is accurate, consistent and relatively fast for potential application in clinical practice.

5. ACKNOWLEDGMENTS

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