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Automatic Left Ventricle Detection in Echocardiographic Images for Deformable Contour Initialization

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Abstract— **The accurate left ventricular boundary detection in echocardiographic images allow cardiologists to study and assess cardiomyopathy in patients. Due to the tedious and time consuming manner of manually tracing the borders, deformable models are generally used for left ventricle segmentations. However, most deformable models require a good initialization, which is usually outlined manually by the user. In this paper, we propose an automated left ventricle detection method for two-dimensional echocardiographic images that could serve as an initialization for deformable models. The proposed approach consists of pre-processing and post-processing stages, coupled with the watershed segmentation. The pre-processing stage enhances the overall contrast and reduces speckle noise, whereas the post-processing enhances the segmented region and avoids the papillary muscles. The performance of the proposed method is evaluated on real data. Experimental results show that it is suitable for automatic contour initialization since no prior assumptions nor human interventions are required. Besides, the computational time taken is also lower compared to an existing method.**

I. Introduction

Ultrasound imaging is an invaluable diagnostic tool in medical analysis. It is important to segment cavities, tissues and organs in an ultrasound image for effective and correct diagnosis. Through accurate boundary estimation in echocardiographic images, cardiologists could then study and assess the different kinds of wall motion abnormalities that is present in the image [1]. Although the cardiac quantifications can be done manually, it is very tedious, and rather difficult to trace these boundaries for each time frame of the cardiac cycle and for large volume of data sets.

To date, there have been many techniques developed for automatically segmenting and tracking the left ventricle [1], [2]. As it has been reported that traditional segmentation algorithms have had limited application, due to the presence of speckle noise in ultrasound images and poor definition of the left ventricle borders, deformable models are generally preferred. However, although deformable models have had good successes in ultrasound applications [3], they still require a good initialization at the beginning of the deformation [4]. Typically, the initialization of the region of interests is outlined manually by the user [3], [5].

In order to automate the initialization process, both semi-automatic [6], [7], [8] and automatic [4], [9] watershed-based left ventricle detection methods have been proposed for ultrasound imaging. In [6], the proposed methods were evaluated on short-axis images, while others used long-axis images [4], [8]. Although Setarehdan and Soraghan [9] proposed an automatic method that was evaluated for both types of images, the assumption that the region of interest is located approximately in the central part of the image plane may not always be valid. Different features of the ultrasound images have also been utilized. In [4], [7] and [8], the temporal properties were used to detect the left ventricular wall. While using the temporal information may provide a better detection, it impedes the online processing of the images.

Hence, there is a need for an automatic and robust approach that is able to detect the left ventricle region in echocardiographic images without human intervention, as well as to allow for online processing. The detected image could then be used as the initialization for the deformable models.

In this paper, a new pre-processing and post-processing stage, coupled with the watershed immersion segmentation method is proposed to fully automate the detection of the left ventricle without the use of temporal information. The pre-processing stage is used to enhance the overall contrast and to reduce speckle noise, while the post-processing stage is used to enhance the detected region to avoid the papillary muscles. Evaluation of the proposed method on real data shows that it is computationally faster and is suitable for automatic contour initialization.

The remainder of the paper is organized as follows. Section II describes the proposed method that includes the pre- and post-processing stages to the watershed segmentation algorithm. Section III analyzes the performance of the proposed method on real data, and Section IV concludes the paper.

II. Proposed Method

Due to suboptimal image quality and the presence of speckle noise, the performance of traditional image segmentation methods tends to be severely affected [1]. Thus, there is a need for pre-processing methods that will both enhance the overall contrast of the ultrasound image, as well as to remove the speckle noise. After pre-processing, the watershed immersion algorithm is applied to extract the

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left ventricle region. Through the combination of the pre-processing stage and the watershed algorithm, a high resolution left ventricle region could be detected. However, as papillary muscles may sometimes appear in the left ventricle chamber, and deformable models favor a convex initialization contour, a post-processing stage is also introduced to enhance the extracted region to avoid the papillary muscles, as well as to ensure the convexity of the region. The final image will then provide an initial boundary of the left ventricle region that is suitable for deformable models.

A. Pre-processing

Low contrast levels and blurring effects are commonly found in ultrasound images [10]. As a result, it is often required to enhance the overall contrast of the image to improve the visibility of object boundaries. A simple, but efficient contrast enhancer, is the histogram equalization method [11]. A point transformation $T(u)$ is specified to make the histogram of the input image more uniform by spreading the pixel intensities evenly in the range $[0, L-1]$, where L is the number of pixel intensity levels. After computing the image histogram of the image, $h(n)$, the cumulative sum, $c(u)$, is computed through

$$c(u) = \sum_{n=0}^{u} h(n), \tag{1}$$

where u is the input pixel value. The transformation $T(u)$ is then given by,

$$T(u) = \frac{(L-1)c(u)}{N_T}, \tag{2}$$

where N_T is the total number of pixels in the image. The histogram equalized image can be obtained by applying the transformation $T(u)$ to the input image.

Through histogram equalization, the overall contrast of the image is increased. However, with the enhancements, speckle noise will also be enhanced. Thus, the median filter is used to reduce the effect of noise. With its edge preserving nature, the median filter replaces pixel u with coordinates (x, y) with a median value of the local $M \times M$ neighborhood,

$$v(x, y) = median\{u(x+i, y+i)\}, \tag{3}$$

where $-W \leq i \leq W$, $v(x, y)$ is the output pixel value and $M = 2W + 1$. In the proposed method, it was empirically found that a 7×7 median filter is suitable for the removal of the speckle noise without severely affecting the quality of the edges.

After enhancing the input image, an image segmentation technique is applied to enhance the chambers in the echocardiographic image. In comparison to histogram thresholding methods, empirical investigation shows that the entropy-based segmentation provides a better separation between the object and background. The optimal threshold value for the entropy-based segmentation method can be determined as follows [12]. Let $\rho_n = h(n)/N_T$. The total entropy of the image H_T and the entropy of each region H_d are given by

$$\begin{aligned} H_T &= -\sum_{n=1}^{N_T} \rho_n \ln \rho_n & (4) \\ H_d &= -\sum_{n=1}^{N_d} \rho_n \ln \rho_n . & (5) \end{aligned}$$

where N_d is the number of pixels in the region. Let $P_u = \sum_{n=1}^{N_d} \rho_n$. The entropies of the background noise H_b and target H_t regions can be calculated as

$$\begin{aligned} H_b(u) &= \ln(P_u) + \frac{H_d}{P_u} & (6) \\ H_t(u) &= \ln(1 - P_u) + \frac{H_T - H_d}{1 - P_u} . & (7) \end{aligned}$$

The optimal threshold is then determined through

$$\tau = \arg\max_u \{H_b(u) + H_t(u)\} . \tag{8}$$

As we are interested in the detection of the echocardiographic chambers, the image is segmented as follows:

$$v = \begin{cases} 0, & u > \tau \\ 1, & u < \tau . \end{cases} \tag{9}$$

In the final step of the pre-processing stage, a distance transform is performed on the binary image, where each pixel is labeled based on its distance to the nearest non-zero pixel. The metric used in the transform is the Euclidean distance:

$$d(m, n) = \sqrt{(x_1 - x_2)^2 + (y_1 - y_2)^2}, \tag{10}$$

where m and n are pixels with coordinates (x_1, y_1) and (x_2, y_2), respectively.

Figure 1 illustrates the processing results of the various steps. As can be observed from the presented results, the pre-processing stage produces a distinct definition of the chambers in the echocardiographic image.

B. Watershed Segmentation

With the success of the watershed segmentation in detecting the region of interests in many medical imaging modalities [1], [2], [5], [13], [14], [15], the watershed immersion algorithm proposed by Vincent and Soille [16] is applied, in this paper, to the gradient image that is produced from the pre-processing stage.

Consider the gradient image obtained from the pre-processing stage as a terrain. Pixels with low gradient values are considered as catchment basins, while those with high gradient values are considered peaks, or watershed lines. The minimum M_i for each catchment basin i is calculated. Progressive flooding of the catchment basins in the gradient image is then performed. When the flooding reaches a height α_i, every catchment basin whose corresponding minimum is smaller than or equal to α_i is assigned the same label. Then, the flooding continues to the next level, $\alpha_i + 1$. If two catchment basins are being merged at level $\alpha_i + 1$, then a dam is built between them to prevent the merger. Therefore, at each level, the labeled catchment basins are extended and new catchment basins are detected. This procedure is

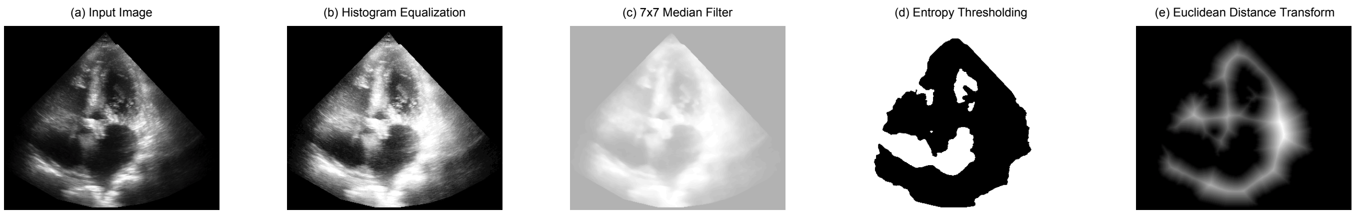

Fig. 1. End result of the four pre-processing steps: (a) input image (b) histogram equalization, (c) median filtering (d) entropy-based segmentation and (e) Euclidean distance transform.

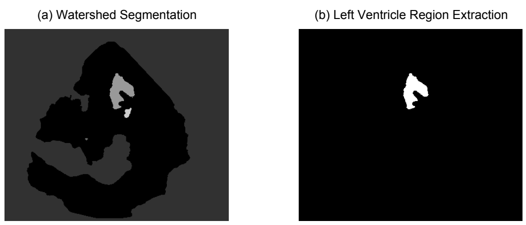

Fig. 2. Segmentation results showing the (a) watershed segmentation and (b) the left ventricle region.

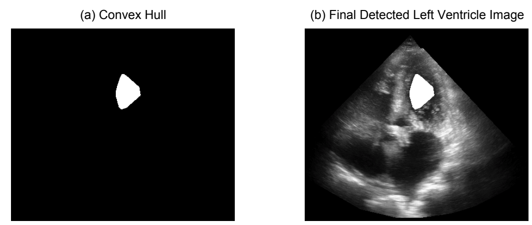

Fig. 3. Post-processing results showing the (a) convex hull and (b) the final detected left ventricle region.

repeated until every pixel in the image has been assigned to a label. The dam built at each stage are the final boundary of the different regions.

As the watershed segmentation may sometimes over-segment the image, its output is then multiplied by the binary image obtained from the entropy-based segmentation to detect the chambers that are present in the image. An image with regions labeled with distinct values is then produced. The second largest region in the whole image is extracted as the left ventricle [17].

It can be observed from Fig. 2 that the left ventricle region extracted from the watershed segmentation corresponds to the actual location of the left ventricle in the input image.

C. Post-processing

As papillary muscles may sometimes appear in the left ventricle chamber, as seen in Fig. 1(a), a post-processing stage is introduced to enhance the extracted left ventricle region to avoid the papillary muscles. In addition, as deformable models tend to favor a convex initialization contour, the post-processing stage also ensures the convexity of the extracted region by performing the convex hull operation [18].

The convex hull of a set of points, S, is the intersection of all convex sets containing S. In other words, the convex hull of S is the smallest convex set that contains all the points. For N points, $p_1, p_2, ...p_N$, the convex hull C is given by

$$C = \sum_{j=1}^{N} \lambda_j p_j, \tag{11}$$

where λ_j are positive weights satisfying $\sum_{j=1}^{N} \lambda_j = 1$.

Generally, the convex hull is calculated as follows. The leftmost and highest point in the whole image is picked as the starting point I. The next point on the hull in the clockwise direction is determined using the cross products. Let P, X and K be the previous point, current point and next point on the hull, respectively. If the cross products of $(X - P) \times (K - P)$ is a negative value, the current X is set as the next point K. This process is iterated though each unused point and is stopped when $K = I$.

By using the convex hull operation, the polygon boundary of the left ventricle region is defined, as shown in Fig. 3, and can then be used as an initialization contour for deformable models.

III. Experimental Results

The proposed method is evaluated on real patient echocardiographic data collected at the St Luke's Hospital, Bethlehem, Pennsylvania. As the apical two-chamber, four-chamber and five-chamber diagnostic views are generally used to assess heart diseases, the proposed method is evaluated on three sample images extracted from the data set.

Figure 4 shows the results of applying the proposed method to the different diagnostic views, in comparison to an existing watershed-based automatic left ventricle initialization method proposed in [4]. It can be observed that the proposed method produces a more refined boundary of the left ventricle region, while the existing method tends to provide a coarser resolution of the boundaries. Thus, the proposed method produces a left ventricle region that is more suitable for deformable models. Besides, the computational time taken by the proposed method is half that of the existing method.

IV. Conclusion

In this paper, we have introduced a new pre-processing stage and post-processing stage to the watershed segmentation algorithm to automatically detect the left ventricle region in echocardiographic images. The detected region can then be used as an initialization contour for deformable models. Experimental results based on real two-dimensional ultrasound data show that the proposed method produces a better result and is computationally faster than the existing method. Thus, it is suitable for automatic contour initialization since no prior assumptions nor human interventions are required.

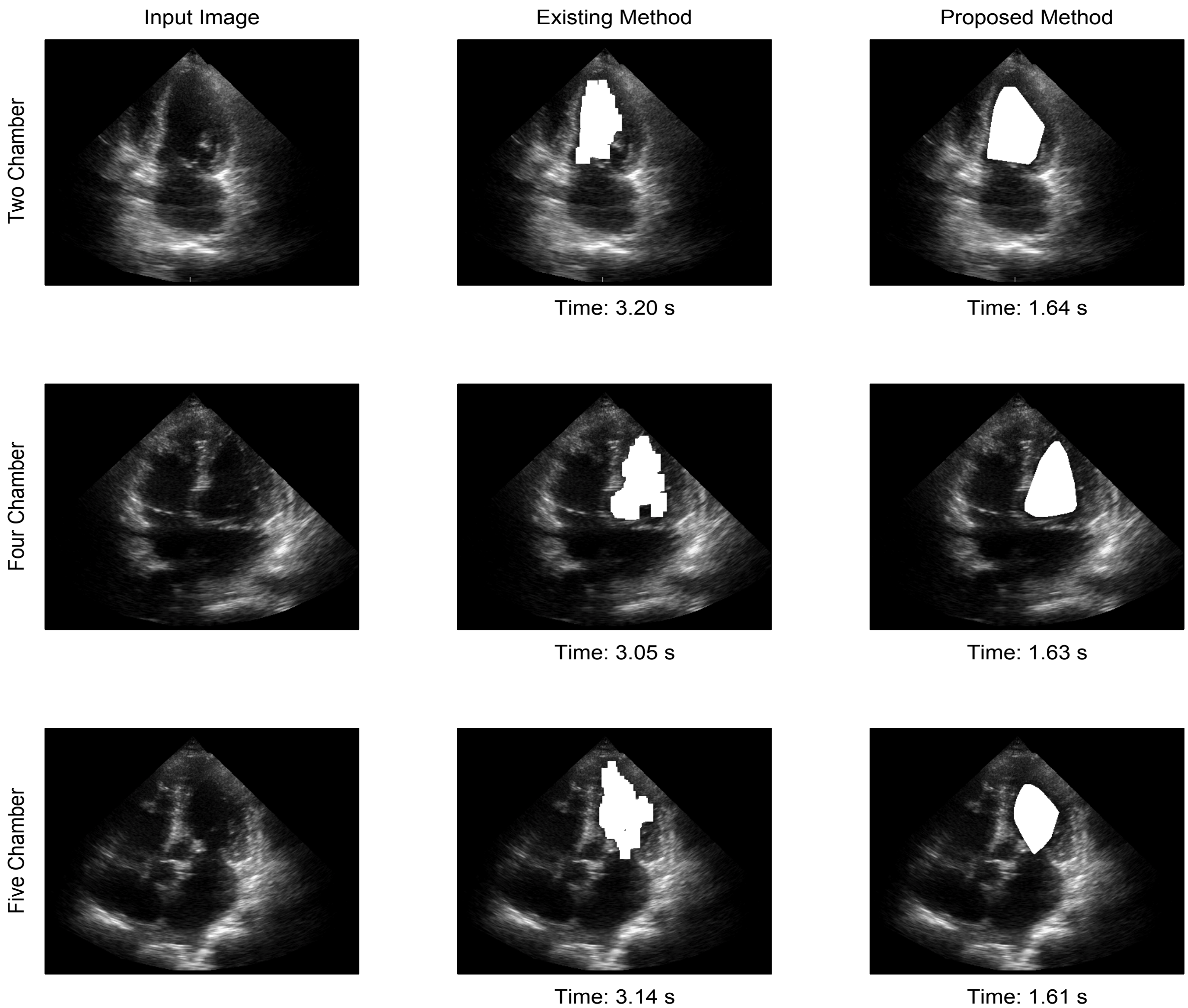

Fig. 4. Evaluation of the proposed method on two, four and five chamber echcocardiographic images.

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