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An Active Contour Based Method for Analyzing Cardiac Quiescence from Echocardiography*

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Abstract— **A semi-automated method for analyzing cardiac quiescence of anatomical cardiac features from two-dimensional echocardiographic cine data is presented. The method utilizes both active contour and optical flow techniques for feature identification and tracking. A curvature-based potential surface was used in the active contour calculations to attract the contour to regions of inflection on the image surface rather than the standard gradient-based surface that attracts the contour to strong edges. After identifying the feature in each frame, the frame-to-frame correlation matrix of the feature was calculated with correlation values corresponding to how well the feature matched between frames. Therefore prolonged regions of high correlation correspond to periods of cardiac quiescence. The location and duration of these periods were automatically identified from the correlation matrix by finding the largest region around each time index with a mean correlation above a specified threshold. In parallel, the position of the feature was calculated for each frame by finding the centroid of the pixel locations inside the contour. From this trajectory, the magnitude of the two-dimensional velocity was calculated. These methods were used to analyze the quiescence of the interventricular septum from an apical four-chamber echocardiogram performed on a human subject. Correlation-derived quiescent phases were observed to coincide with periods of the cardiac cycle with minimal velocity magnitude.**

I. Introduction

This work is motivated by the need to identify quiescent periods of the heart within the cardiac cycle. These periods of quasi-stationarity are critical to gate advanced cross-sectional imaging modalities, such as computed tomography (CT) and magnetic resonance imaging (MRI). Reconstructed image quality is highest when acquisition occurs during phases with minimal motion. Cardiac motion during acquisition will result in blurring and lower diagnostic quality images. Currently, triggering of data acquisition for CT and MRI cardiac imaging almost exclusively relies on the electrocardiogram (ECG), which is a surrogate marker for cardiac motion [1]. Although the ECG signal is highly correlated with cardiac motion, it is an indirect representation of cardiac state because it only indicates the instantaneous electrical activity of the heart. On the other hand, echocardiography permits direct observation of the heart, providing real-time motion information. For this reason, echocardiography has the potential to more accurately identify acquisition time windows for CT and MRI [2]. With this in mind, a method for identifying these gating windows for a specific cardiac feature was designed.

The method developed extracts motion information for a specific cardiac feature identified by applying active contour techniques to echocardiography data. The optical flow of the echocardiography sequence is used to aid initialization of the active contour for each frame after the first.

The rest of this paper is organized as follows. A brief background on active contours and optical flow is provided in Section II. Section III presents the methods we employ for feature identification and quiescent phase analysis. Preliminary quiescent phase statistics and results for a human subject are shown in Section IV, and conclusions are presented in Section V.

II. Background

Active contour methods provide a robust and flexible method for feature identification in an image [3]. The active contour itself is defined as a deformable curve

$$\mathbf{v}(s) = \langle x(s), y(s) \rangle, \tag{1}$$

where s is the contour length, normalized to one. The active contour problem is solved by minimizing the energy function in

$$E_{contour} = \int_0^1 \left(E_{int}(\mathbf{v}(s)) + E_{img}(\mathbf{v}(s)) \right) ds, \tag{2}$$

where $E_{int}(\mathbf{v})$ is the internal energy associated with the geometry of the contour and $E_{img}(\mathbf{v})$ is the image energy determined by image characteristics along the contour. A contour solution is found by minimizing the combination of these energies, balancing the constraints on the contour shape and the influence of the image on the contour.

The optical flow of an image sequence is an estimate of the motion between frames. This motion estimation can be found for a given pixel by defining a small neighborhood around that pixel in the first frame and correlating that neighborhood across a search range in the second frame. In this manner, a motion vector for each pixel can be calculated. By performing motion estimation on each frame, motion vectors for a complete sequence can be calculated.

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III. METHODS

Echocardiography, ultrasound of the heart, was chosen as the modality for quiescent phase identification because it is real-time, has high temporal resolution, is noninvasive, and does not require ionizing radiation. Quiescence was determined from a sequence of B-mode, or two-dimensional, echocardiography frames. Echocardiography and ECG data were recorded synchronously. The ECG signal was used as the reference time frame and marks the beginning and end of the cardiac cycles.

A. Active Contours for Feature Identification

The cardiac feature-tracking algorithm identifies the feature of interest in each frame of echocardiographic data. The algorithm is initialized with the user providing a rough outline of the cardiac feature in the first frame of the sequence. Active contour techniques are used to solve for the outline of the feature in each frame. For all frames after the first, the position of the previous contour along with optical flow data of the sequence are used to initialize the active contour algorithm.

Active contour feature segmentation closely follows the method presented in [4] with the algorithm setup reproduced below for completeness. The contour $\mathbf{v}(s)$ is discretized as a sequence of M vertices as

$$\mathbf{X} = \begin{bmatrix} x_0 & y_0 \\ x_1 & y_1 \\ \vdots & \vdots \\ x_{M-1} & y_{M-1} \end{bmatrix}. \quad (3)$$

The energy function in (2) can be minimized by recasting the problem as a physical model with the contour having properties such as weight and stiffness, corresponding to E_{int}. The contour can be seen as being placed on a potential surface derived from the image. In this sense the contour has potential energy, corresponding to E_{img}. The motion of this model is defined as

$$\mathbf{M}\ddot{\mathbf{X}}(t) + \mathbf{C}\dot{\mathbf{X}}(t) + \mathbf{K}\mathbf{X}(t) = \mathbf{F}(t), \quad (4)$$

where $\mathbf{M} = \mu \cdot \mathbf{I}_M$, with μ the mass assigned to each vertex of $\mathbf{X}$ and $\mathbf{I}_M$ the $M \times M$ identity matrix, $\mathbf{C} = \gamma \cdot \mathbf{I}_M$, with γ the constant damping factor, $\mathbf{K}$ an $M \times M$ stiffness matrix for the contour, and $\mathbf{F}$ the matrix of image force vectors at each vertex of $\mathbf{X}$. Curve evolution was performed using the method presented in [4, 5].

The most common potential surface is derived from the negative magnitude of the squared gradient of the image intensity,

$$E_{img} = -\|\nabla I\|^2. \quad (5)$$

This potential surface will be smallest near points with large gradient magnitude, i.e., strong edges.

The active contour method for this work departs from the gradient based potential surface used in [4]. The potential surface applied here is derived from the squared mean curvature at each pixel of the image,

$$E_{img} = \mathcal{K}(I)^2, \quad (6)$$

where $\mathcal{K}$ is the mean curvature. This surface was chosen to attract the contour to regions of the image where the mean surface curvature is zero, corresponding to inflection points on the surface. It was observed that this choice of surface led to more stable contour evolution for features in regions of low image intensity that may not have a large gradient magnitude.

The active contour algorithm is initialized by providing an outline of the cardiac feature of interest in the first frame. The active contour is then evolved until a solution to (2) is reached. Let $\widetilde{\mathbf{X}}_n$ be the initial estimate of the contour location for frame n prior to any active contour calculations and $\mathbf{X}_n$ be the solved contour for frame n. Then, $\mathbf{X}_1$ identifies the feature in the first frame. From the optical flow, the motion vectors associated with the vertices of $\mathbf{X}_1$ are used to provide an initial estimate of the location of the contour in the second frame, $\widetilde{\mathbf{X}}_2$. Contour initialization for $n > 1$ can be summarized as

$$\widetilde{\mathbf{X}}_n = \mathbf{X}_{n-1} + d\mathbf{X}_{\boldsymbol{n-1}}, \quad (7)$$

where $d\mathbf{X}_{\boldsymbol{n}}$ is a matrix of the x and y components of the motion vectors corresponding to the contour vertices in $\mathbf{X}_n$. The optical flow of the sequence was computed using Singh's method presented in [6].

Active contour feature identification is then performed for an entire sequence of B-mode echocardiography data with the output being the sequence of $\mathbf{X}_n$.

B. Correlation Techniques to Identify Quiescent Phases

Quiescent periods of the cardiac cycle can be identified by computing the frame-to-frame correlation of the feature of interest. The premise of this method is that the correlation among frames will be high during periods of quiescence because the feature will be quasi-stationary. During periods of large cardiac motion the correlation will be low because the feature will not align between frames.

Let S_n be the set of all the pixel locations inside $\mathbf{X}_n$. Prior to correlation calculations, the non-feature portion of each frame is set to zero. The feature correlation between two such frames, I_i and I_j, is then defined as

$$C(i,j) = \frac{\sum_{(x,y)\in S}\left(I_i(x,y) - \bar{I}_i^S\right)\left(I_j(x,y) - \bar{I}_j^S\right)}{\sqrt{\sum_{(x,y)\in S}\left(I_i(x,y) - \bar{I}_i^S\right)^2 \sum_{(x,y)\in S}\left(I_j(x,y) - \bar{I}_j^S\right)^2}}, \quad (8)$$

where $S = S_i \cup S_j$ and $\bar{I}_i^S$ is the mean of the pixel intensities of I_i in S. This choice of S penalizes the correlation for mismatch between S_i and S_j, while still promoting matching intensity values in $S_i \cap S_j$. Cardiac quiescence is indicated by square regions of high correlation along the diagonal of the correlation matrix, $C(i,j)$. The correlation matrix for one cardiac cycle is plotted in Fig. 1 (a).

The process of calculating the timing and duration of these phases was automated. For each point on the diagonal of $C(i,j)$, the mean correlation of a square neighborhood centered at that point is calculated. The size of this neighborhood is increased until the mean is no longer above a specified threshold. Here quiescence is defined as the

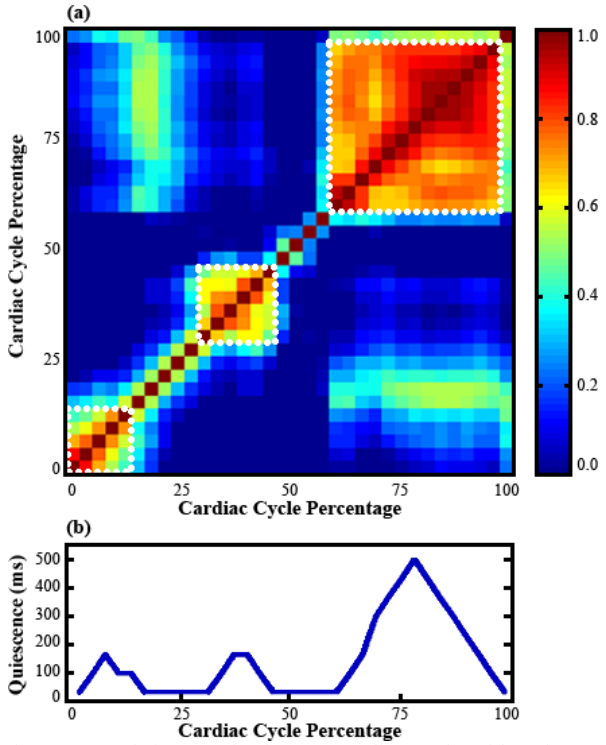

Fig. 1. Correlation matrix for one cardiac cycle with quiescent regions with mean correlation greater than 0.7 outlined (a) and the corresponding quiescence plot (b). Here quiescence refers to the width in milliseconds of the square region at each time index.

duration of the gating window indicated by the width of largest square region along the diagonal of the correlation matrix with mean correlation above some specified threshold. The quiescence plot depicting the quiescence at each time index in Fig. 1 (a) is shown in Fig. 1 (b). A threshold value of 0.7 was used, and periods of quiescence are also indicated in Fig. 1 (a) by dotted lines. For each cycle, the optimal gating window position and duration can be identified by finding its peak quiescent value.

C. Position and Velocity Estimation

The two-dimensional position of the contour is calculated for each frame by finding the average location of the pixels inside the contour, defined as

$$\bar{x}_n = \frac{1}{N}\sum_{x \in S_n} x\,, \quad \bar{y}_n = \frac{1}{N}\sum_{y \in S_n} y, \tag{9}$$

where S_n is the set of all pixel locations inside the contour $\mathbf{X}_n$ and N is the number of elements in S_n. A composite motion trajectory is then formed by averaging multiple cycles of $\bar{x}_n$ and $\bar{y}_n$, segmented using the R-peak of the ECG signal as the start reference for each cardiac cycle.

The two-dimensional velocity of the feature is calculated by taking the first difference of the composite position signal. The magnitude of the velocity is then used as an indication of quiescence for the heartbeats over which the composite signal was formed with periods of small velocity magnitude corresponding to quiescent phases.

D. Preliminary Human Study

B-mode echocardiographic data was obtained in the apical four-chamber view from a 23 year-old female with no known cardiac conditions. Approval for this study was provided by the Emory University Institutional Review Board. Synchronous B-mode and ECG data were recorded at rates of 30 frames-per-second and 200 Hz, respectively, using a SonixTOUCH Research ultrasound machine (Ultrasonix, Vancouver, BC, Canada). The frame size for processing was 240-by-320 pixels.

IV. Results

The active contour algorithm was initialized by manually providing an outline of the interventricular septum in the first frame of a B-mode sequence. The interventricular septum was tracked throughout the sequence for 33 cycles with an average heart rate of 55 beats per minute. An example of the contour, $\mathbf{X}$, from one frame is provided in Fig. 2. The correlation matrix, $C(i,j)$ defined in (8), was calculated and the largest quiescent phase for each cardiac cycle was identified using a correlation threshold value of 0.7. A portion of the quiescence plot and the corresponding gating windows indicated on the ECG are shown in Fig. 3 (a) and (b) respectively. The statistics of the optimal gating windows for each cycle are provided in Table I.

TABLE I. Quiescent Phase Statistics

Number of Cardiac Cycles	Mean Heart Rate	Mean Window Position	Mean Window Length
33	55 +/- 3.2 bpm	75 +/- 2.5 %	399 +/- 75 ms

Lastly, the position and velocity of the septum were calculated using each frame's contour. The composite x and y position trajectories formed by averaging each cycle are plotted in Fig. 4 (a) and (b). From these signals the x and y velocities were calculated with their magnitude shown in Fig. 4 (c). A choice of 2.0 cm/s as a velocity magnitude threshold for quiescence was found to agree with the correlation derived window statistics from Table I. This threshold results in a gating window centered at 76.5% with a duration of 365 ms, shown as the shaded region in Fig. 4 (c).

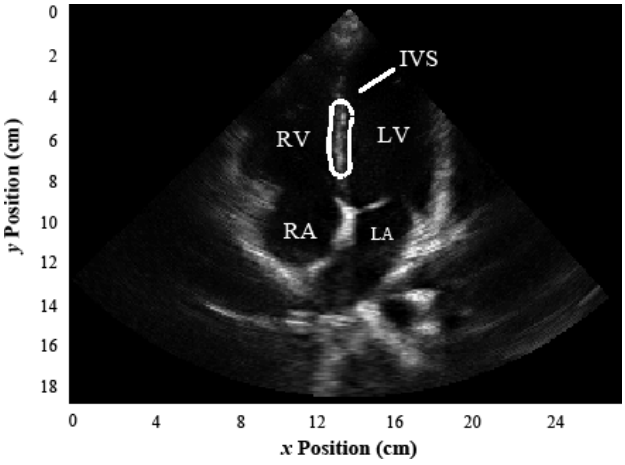

Fig. 2. B-mode frame obtained from an apical four-chamber view of the heart, with solved contour shown around the interventricular septum (IVS). The four chambers: right ventricle (RV), right atrium (RA), left ventricle (LV), and left atrium (LA) are shown.

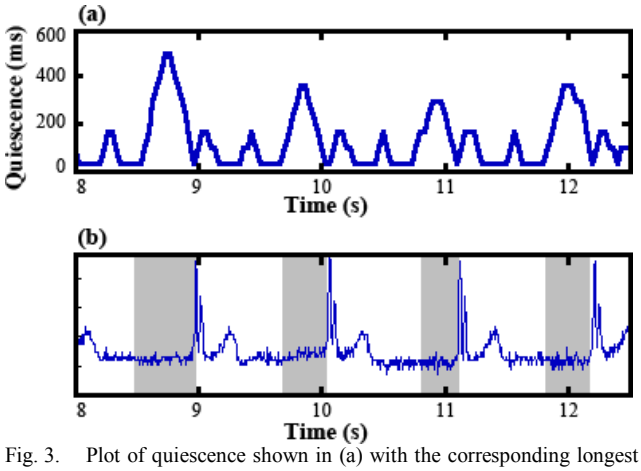

Fig. 3. Plot of quiescence shown in (a) with the corresponding longest gating windows shaded for each cycle on the ECG in (b).

V. Discussion and Conclusion

A method was presented for analyzing echocardiographic images to identify the quiescent phases associated with a cardiac feature within a cardiac cycle. Accurate knowledge of these phases will allow for further study of optimal gating windows for advanced cross-sectional imaging modalities such as CT and MRI. Improved gating can yield higher quality images in both CT and MRI. An added benefit of more accurate gating in CT is a reduction in X-ray tube current ON time and consequently a reduction in patient radiation dose.

The proposed method determines quiescence based on two-dimensional motion analysis. This is a benefit over traditional Doppler techniques for motion quantification, which provides velocity estimation based on the Doppler shift of the returned ultrasonic pulse. Doppler shift is only sensitive to the motion component in the direction parallel to the ultrasound beam. Thus, the accuracy of Doppler velocity estimates degrades rapidly as the direction of the motion moves away from that of the ultrasound beam. Our method avoids this issue by analyzing the two-dimensional motion of the cardiac feature in B-mode frames.

A further application of this work is to analyze the variation in gating windows at different heart rates and among different individuals. It has been demonstrated that cardiac quiescence varies for different heart rates and from subject to subject [2, 7]. The proposed method will allow us to investigate this further and to potentially derive patient-specific gating parameters. A patient could be examined prior to receiving a cardiac CT or MRI scan to determine a personalized gating protocol.

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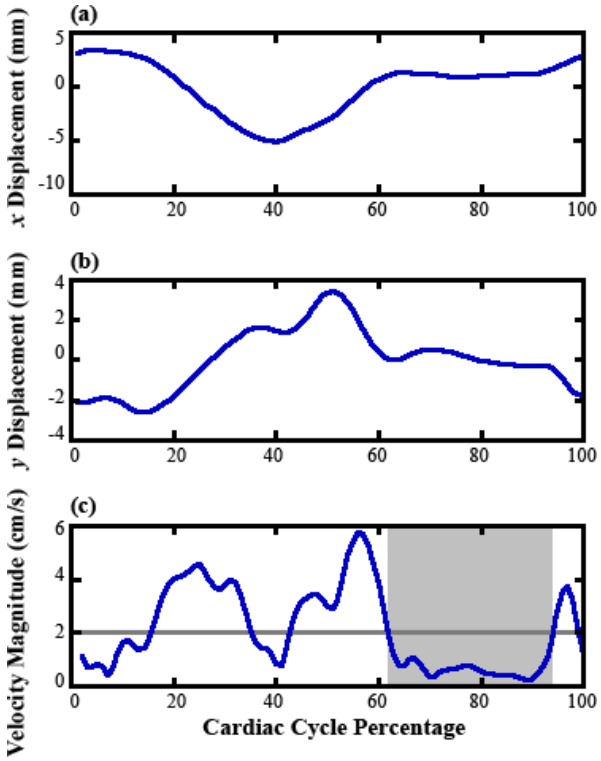

Fig. 4. Vertical displacement (a), horizontal displacement (b), and velocity magnitude (c) of the active contour centroid plotted as a function of cardiac cycle percentage. Longest region with velocity magnitude less than 2.0 cm/s is shaded in (c).

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