CARDIAC WALL TRACKING USING DOPPLER TISSUE IMAGING (DTI).

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ABSTRACT

Doppler Tissue Imaging (DTI) is a new technique that provides an image information on values of intramyocardial wall velocity. This specific capability makes easier the automatic myocardial boundary extraction and tracking task. We apply a variation on active contour models to an M-Mode Ultrasound image. This is in fact an image made from a time sequence of digitized 1D images. To the feature energy term derived from the grey scale conventional ultrasound image, is added a velocity term that makes the time displacement in the image close to the DTI data velocity. This permits data fusion of the two images obtained from conventional and DTI techniques. We illustrate the efficiency of the method that is now used by clinicians.

1. INTRODUCTION

The analysis of cardiac deformations has given rise to a large amount of research in medical image understanding. Indeed, cardiovascular diseases are the first cause of mortality in developed countries. Although various imaging techniques make it possible to get dynamic sequences of high resolution 3D images, they are not yet used for routine diagnosis by physicians. One particularity of our work is that it was made in the hospital with PC or Macintosh for immediate use by clinicians. This is why we were interested in the analysis of Ultrasound images since they are easy to obtain for the cardiologist and less impressive for the patient.

We present some results obtained using active contours with the brand new technique of Doppler Ultrasound imaging. Conventional 2D Echocardiography allows the cardiologist to make some diagnosis by visualization of the deformation of the cardiac walls using his expert analysis to detect if they behave correctly. Doppler images give a velocity information of the blood flow in and out the ventricles cavities. They also help for the diagnosis of valvular and myocardial diseases, involving pathological blood flow. The new technique used here, called Doppler Tissue Imaging (DTI), gives a visualization of the velocity of the intramyocardial wall motion instead of the blood flow velocity. Figure 1 shows a bidimensional ultrasound image together with its DTI mode image. The frame rate and the temporal resolution of 2D images do not allow the study of myocardial wall velocities over a whole cardiac cycle. Therefore, in their regular practice, cardiologists need "M-mode" images.

These are obtained by choosing a 1D segment on the image and visualize its evolution in time as a 1D+t image where each column correspond to the same 1D slice at a given time. Figure 2 shows an M-mode image together with its DTI mode.

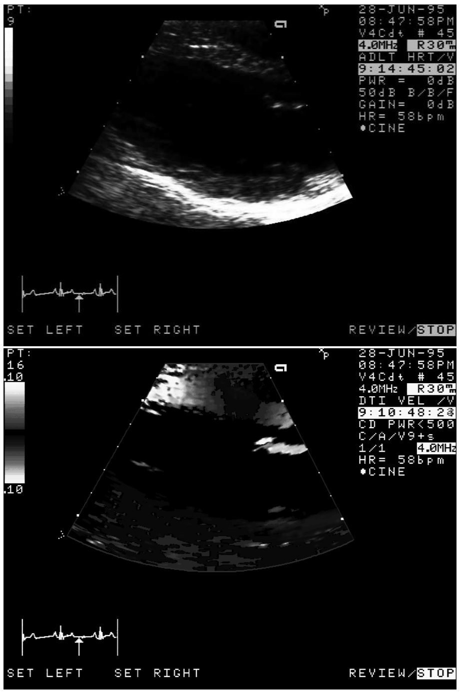

Figure 1: Example of echocardiographic Bidimensional image : above, grey scale image, below, Doppler Tissue Image in velocity mode

Since the DTI image is given as a color image with a color scale on the left side reflecting the amplitude and direction of the velocity, a first job was to convert it into an image of velocity by analysis of the image and projection on the closest color. This was done by

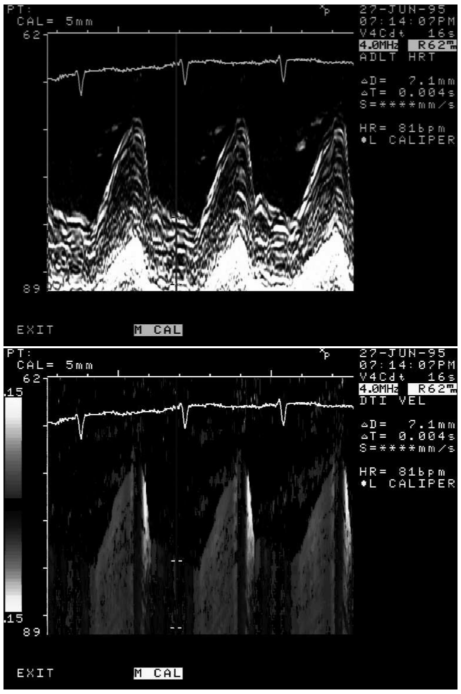

Figure 2: Example of Left Ventricular Posterior Wall M-Mode Ultrasound image : above, grey-scale image, below, color Doppler Tissue Image

finding for each pixel of the image, which color is the closest by Euclidean distance on the look up table, on the left of the image. This procedure is detailed in [1].

2. ACTIVE CONTOURS

Much work using deformable models have studied spatial segmentation and tracking of the ventricle boundaries on classic ultrasound images to make the job of the physician as much automatic as possible. Already in the first work on snakes [2], an application was shown for tracking of lips in a face image. This idea was used for spatial tracking in a stack of 2D images to make 3D reconstruction from cross sections in [3, 4]. Each model is initialized by the solution in the previous slice. Another important work on tracking was presented in [5]. Segmentation for ultrasound images was presented as an application of the balloon model in [3]. Temporal tracking in a time sequence of ultrasound images was studied in [6, 7]. The tracking is helped by a Kalman filter in the Kalman snakes [8]. Other references can be found in the book [9]. In [10, 11], the initialization at each step is given by the previous solution deformed by prediction in the direction of the local flow obtained by optical flow analysis of the sequence of images. Our goal is to follow the deformation of a wall by including in the energy of the model the knowledge of its local velocity.

The contour model, as introduced in [2], is a curve $C(s) = (x(s), y(s))$ that minimizes an energy of the following form:

$$E(C) = \int_\Omega w_1 \|C'(s)\|^2 + w_2 \|C''(s)\|^2 + P(C(s))ds$$

where P is the potential associated to the external forces.

If C is a local minimum for E, it satisfies the associated Euler-Lagrange equation:

$$\left\{ \begin{array}{l} -(w_1 C')' + (w_2 C'')'' = -\nabla P(C) = F(C) \\ C(0), C'(0), C(1) \text{ and } C'(1) \text{ given.} \end{array} \right. \quad (1)$$

As explained in [3, 4], the energy is not convex, because of the potential term and there may be many local minima of E. To find a good contour in a given area, we suppose we have a rough estimate of the curve. We impose the condition to be "close" to this initial data by solving the associated evolution equation [3]. When discretized by finite elements, the following iterative scheme is used to solve equation (1):

$$(Id + \tau A)\mathcal{C}^{n+1} = (\mathcal{C}^n + \tau F(\mathcal{C}^n)) \quad (2)$$

where $\mathcal{C}^n$ is the vector of nodes at iteration n.

3. VELOCITY ACTIVE CONTOURS

We have used a simplified active contour to track the cardiac wall in these M-mode images. Since the x dimension of the image represents time, we can follow the motion of a point on the cardiac wall with time. This is easy to see for example on the border of the wall in the top image of figure 2. The cavity is in black and the contour between the cavity and the bottom region corresponds to the time evolution of one point on the internal border of the cardiac wall. We can see the same kind of curves by moving this boundary downwards. These curves provide the clinician information on the deformation of the wall during a cardiac cycle. We use a deformable model to extract these curves. A curve represents in fact the temporal tracking of one point on the cardiac wall. It is then natural to parametrize the curve by time, this means that we look for a curve $C(t) = (t, y(t))$. It is initialized first by an easy guess, preferably a segment, and it deforms according to energy minimization (see figures 3 and 4). The energy is composed of classic regularization terms and external terms are obtained from the gradient image as well as the DTI image:

$$\begin{aligned} E(y) = \quad & \int_0^T \omega_1 \left|y'(t)\right|^2 + \omega_2 \left|y''(t)\right|^2 \\ & + P_{edge}(C(t)) + P_{velocity}(C(t))dx \end{aligned} \quad (3)$$

where P_{edge} attracts the curve to edges and $P_{velocity}$ constrains the curve to get $y'(t)$ close to the given velocity measured by the DTI. This makes a fusion of the information contained in the two image modalities, grey scale conventional ultrasound and DTI. As introduced and used in [3, 4], the potential is derived from

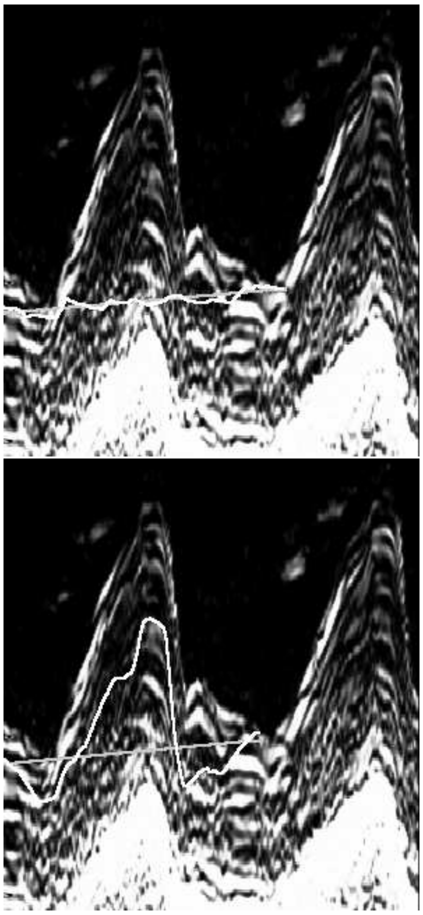

Figure 3: Advantage of using the DTI mode: above, the potential is based only on the grey scale image; below, we obtain a good result since the DTI velocity information is included in the model's energy. In both images, the initialization is shown in grey, the final curve in white.

a set $\mathcal{S}$ of already extracted **edge points** from the grey scale ultrasound image. The potential we used is a smoothed version of this binary image of edges. It is obtained by convolution with a Gaussian kernel. A distance map could also be used.

The DTI image gives a value $v(x, y)$ of the velocity in the direction of the radius from the transducer. In the M-mode image, this direction is always vertical. The motion is one dimensional. The velocity term is $|y'(t) - v(C(t))|^2$. This can be replaced by a weaker constraint where only the direction of motion is the same as the one given by v. This means we constrain the point to go either up or down. In the bottom image of figure 2, we see that in each cycle there is one part where the velocity is positive and a second where it is negative.

When we discretize with t, according to the pixel size, assuming the time step is 1, the constraint means

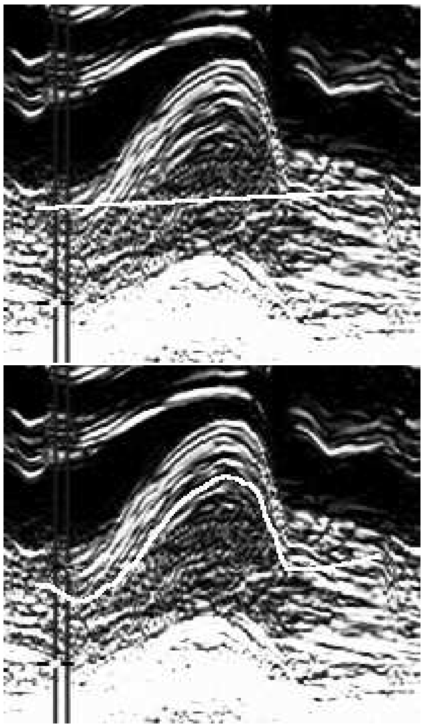

Figure 4: Another example using DTI in velocity mode: initialization above, solution below.

that we should have:

$$y_{t+1} = y_t + v(t, y_t) \tag{4}$$

In the discretized energy, we have two terms that contain a given y_t:

$$\begin{aligned} E_{velocity}(t, y) &= (y_{t+1} - y_t - v(t, y_t))^2 \\ &+ (y_t - y_{t-1} - v(t-1, y_{t-1})^2 \end{aligned} \tag{5}$$

The force applied to the curve at point $(t, y(t))$ is thus:

$$F_{velocity}(t, y) = -\frac{\partial E_{velocity}}{\partial y_t} \tag{6}$$

$$\begin{aligned} &= 2((y_{t+1} - y_t - v(t, y_t))(1 + \frac{\partial v}{\partial y_t}(t, y_t)) \\ &- (y_t - y_{t-1} - v(t-1, y_{t-1}))) \end{aligned} \tag{7}$$

where we can neglect the term $\frac{\partial v}{\partial y_t}$ with respect to 1. Similar to (2), the iterative scheme obtained for y at iteration n is:

$$\begin{aligned} y_t^{n+1} &- y_t^n = \tau\alpha(y_{t-1}^{n+1} - 2y_t^{n+1} + y_{t+1}^{n+1}) \\ &- \tau\beta(y_{t-2}^{n+1} - 4y_{t-1}^{n+1} + 6y_t^{n+1} - 4y_{t+1}^{n+1} + y_{t+2}^{n+1}) \\ &+ \tau(F_{velocity}(t, y^n) + F_{edge}(t, y^n)) \end{aligned} \tag{8}$$

which gives in matrix form:

$$(I_d + \tau A)Y^{n+1} = (Y^n + \tau F(Y^n)) \tag{9}$$

To get it to work in real time on a small machine, the implementation was done by explicit finite differences to avoid the matrix inversion [4]. To get faster convergence, we used an SOR algorithm [12]. This means we have:

$$\begin{aligned}(1+ \quad 2\tau\alpha \quad & +6\tau\beta)y_t^{n+1} = & (10)\\ & y_t^n + \tau\alpha(y_{t-1}^{n+1} + y_{t+1}^n) & (11)\\ - \quad & \tau\beta(y_{t-2}^{n+1} - 4y_{t-1}^{n+1} - 4y_{t+1}^n + y_{t+2}^n) & \\ + \quad & \tau(F_{velocity}(t, y^n) + F_{edge}(t, y^n)) &\end{aligned}$$

We used free boundary conditions at the end points. In our examples, we used the values $\tau = 0.1, \alpha = 1, \beta = 0.5$. In examples of figures 3 and 4, we see the advantage of using the velocity term in the energy. With this term, we get a good result with a simple initialization. In practical use of the method, initialization is provided by hand, keeping the spirit of interactivity of the initial work on snakes [2]. However, it could also be initialized by horizontal segments.

Results were satisfying compared to manually traced curves by the cardiologist. A direct medical application of these results was to show quantitatively that velocity of the internal wall (endocardium) is higher than for the external wall (epicardium) [13, 14].

The computerized analysis of "M-mode" images is now available with a PC in the echo-lab. The generalization of the method for a time sequence of 2D images, tracking the complete boundary of the wall, is straightforward. It needs however further study or waits for more powerful PC-like computers.

4. REFERENCES

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