Semi-automatic noninvasive assessment of local myocardial motion Using M-mode Echocardiogram

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Abstract—**we propose a new methodology for motion tracking of local myocardial tissue on M-mode echocardiograms. This methodology is applicable to the quantitative assessment of myocardial performance in clinics. The M-mode echocardiogram is widely used in clinics to measure diagnostic indexes like thickening and thinning of myocardial muscle layers. To measure such indexes, doctors are required to track myocardial motion manually, however the tracking of myocardial motion by hand is tedious and time-consuming process. Our proposed method is able to track the myocardial motion on M-mode echocardiograms automatically by employing DP-based optimization. In this report we present the proposed method.**

Keywords— **Myocardial Motion Tracking, Ultrasonic Imaging, Echocardiography, Dynamic Programming**

I. INTRODUCTION

Echocardiography is an essential and necessary tool for diagnosis and treatment of cardiovascular diseases. The use of echocardiograms in clinical cardiology serves clinicians and doctors with multifold advantages including noninvasiveness of diagnosis and real-time imaging. In clinical cardiology, the analysis mainly relies on visual inspection or manual measurements by experienced cardiologists. Manual methods are tedious and time consuming, and visual assessment leads to qualitative and subjective diagnoses that suffer from a considerable inter- and intra-observer variability. Therefore, an automated computer-based analysis is highly desirable to obtain more objective and quantitative diagnoses.

Fig.1 shows an example of clinical echocardiogram. In the figure, $(a)$and (b) are B-mode and M-mode echocardiograms respectively. M-mode echocardiogram is widely used in clinics to measure diagnostic indexes like thickening and thinning of myocardial muscle layers. It is a diagnostic ultrasound presentation of the temporal changes in echoes in which the depth of echo-producing interfaces is displayed along vertical axis and time is displayed along the horizontal axis, recording motion of the interfaces toward and away from the transducer.

Several approaches have been proposed to quantify the motion of myocardium from two dimensional (2-D) echocardiograms [1-4]. However, in these approaches, the spacial and temporal resolutions of quantification and tracking of myocardial motion is strongly depends on

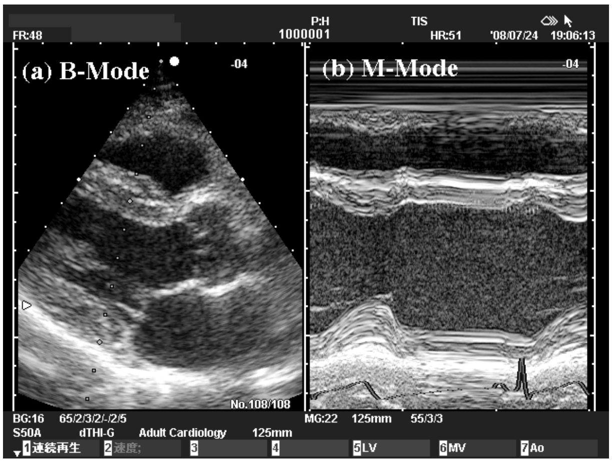


Fig.1 Example of clinical echocardiogram: (a)B-mode and (b) M-mode echocardiograms

those of imaging performance of ultrasonic equipment. Particularly, echocardiograms employ low-pass filters on ultrasonic signals in the image generation process. Consequently, the special resolution of images becomes low and insufficient for detailed diagnosis of local, inner myocardial tissues. To overcome this low temporal resolution problem, several approaches employ raw ultrasonic signal from which echocardiograms are generated [5-10].

In this paper, we propose a novel approach to track local (inner wall) myocardium. The proposed approach employs M-mode ultrasonic Doppler signal for velocity estimation and Dynamic Programming (DP) based motion tracking method. DP is a method of solving problems exhibiting the properties of overlapping sub-problems and optimal substructure that takes much less time than other methods in mathematics and computer science. In the method, DP is employed to find a trajectory which minimizes the objective function.

II. MYOCARDIAL MOTION TRACKING

We describe here how to track the local myocardial motion using the proposed DP tracking method. The method consists of two main stages, as shown in Fig.2. At the first step, velocity field in the myocardial wall is estimated by the correlation weighted mean algorithm [10]. After that, in the motion tracking stage, instantaneous displacement on a set of tracking points is calculated from the estimated velocity field and the DP tracking method tracks the motion of the tracking points.

A. Acquisition system for M-mode ultrasonic Doppler signal

To assess inner-wall myocardial motional performance, myocardial local motion should be tracked with high temporal resolution. For instance, typical normal left ventricular myocardium has a thickness of 10 millimeters, so a temporal resolution of an order of sub-millimeter is required.

We employ M-mode ultrasonic Doppler signal to achieve motion tracking with such the high temporal resolution. Fig.2 illustrates the block-diagram of signal acquisition system in this research.

An ultrasonic transducer placed on the chest transmits pulsed ultrasonic toward the heart, and receives the back-scattered ultrasonic. The transducer converts the received ultrasonic to electric signal $y(t)$. The electric signal $y(t)$ is quadrature demodulated and then A/D converted to be imported to PC as discrete Doppler signal $\boldsymbol{Z}$. Discrete Doppler signal $Z = \{z(x_{i,}t_k)\}$ consisting of sine component $z_s(x_i, t_k)$ and cosine component $z_c(x_i, t_k)$ is represented by complex expression as follows,

$$z(x_i, t_k) = z_s(x_i, t_k) + jz_c(x_i, t_k), \quad (1)$$

$$x_i = ic_0T_s,$$

$$t_\mathrm{k} = \mathrm{k}\Delta\mathrm{T},$$

where x_i and t_k are the distance from ultrasonic transducer to i-th sampling point and the time when k-th ultrasonic plus is transmitted respectively. The constant c_0 is sound speed in human body, i.e. 1530 [m/s]. T_s and ΔT are sampling period and repetition period of the ultrasonic pulse respectively, and j is the imaginary unit.

Fig.4 examples an actual ultrasonic Doppler signal at t_k and t_{k+1}.

B. Velocity measurement for myocardium

Instantaneous velocity of each sampling point $v(x_j, t_k)$ is first derived from the discrete Doppler signal in the same way as in the conventional method [6].

$$\theta(x_j, t_k) = \tan^{-1}\frac{z_c(x_j,t_k)}{z_s(x_j,t_k)}, \quad (2)$$

$$\Delta\,\theta(x_j, t_{k+1}) = \theta(x_j, t_{k+1}) - \theta(x_j, t_k) \quad (3)$$

$$v\,(x_{j,}t_k) = \frac{\mathrm{c_0\Delta\theta(x_j,t_k)}}{4\pi f_0} \quad (4)$$

where f_0 is the centroid frequency of the ultrasonic pulse. Since the obtained velocities generally contain some error due to speckle noise, the errors are accumulated in

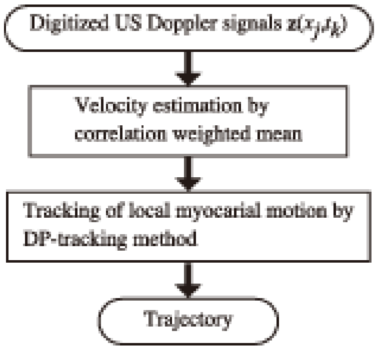


Fig. 2 The flow of myocardial local motion tracking process

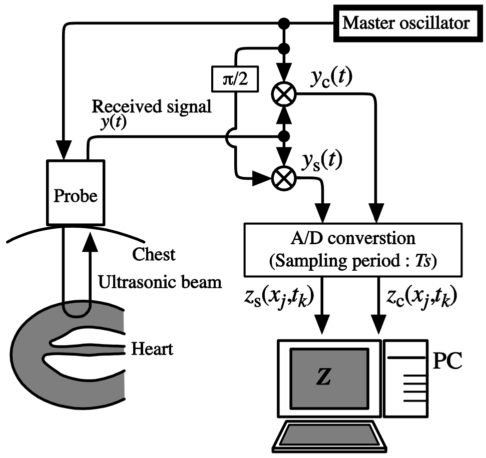


Fig. 3 Block-diagram of data acquisition system

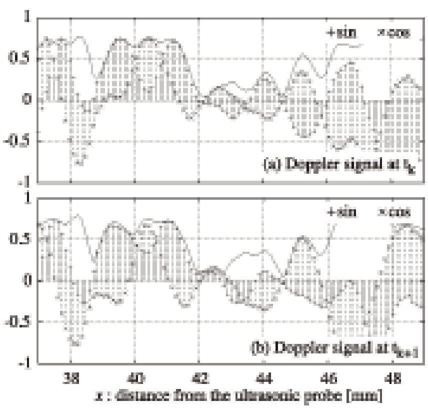


Fig.4 Example of acquired ultrasonic Doppler signal

the tracking process, and severely deteriorate the accuracy of the motion tracking.

The proposed method reduces these errors by taking correlation weighted mean of the velocities at several sampling points. The correlation weighted mean velocity $v'(x_j, t_k)$ at a tracking point x_j is defined by

$$v'(x_j,t_k) = \frac{\sum_{l=0}^{N-1}\gamma(x_{j+l},x'_{j+l},t_k)v(x_{j+l},t_k)}{\sum_{l=0}^{N-1}\gamma(x_{j+l},x'_{j+l},t_k)}\ , \tag{5}$$

where N is length (sample size) of ultrasonic pulse, and $\gamma(x_{j+l}, x'_{j+l}, t_k)$ is the correlation of the magnitude around the sampling point before and after a step of motion tracking , which is defined by,

$$\gamma(x_{j+l}, x'_{j+l}, t_k) = \frac{|z_{j,k}|\bullet|z'_{j,k+1}|}{\sqrt{|z_{j,k}|\bullet|z_{j,k}|}\sqrt{|z'_{j,k+1}|\bullet|z'_{j,k+1}|}}\ , \tag{6}$$

$$|z_{j,k}|\bullet|Z'_{j,k+1}| =$$

$$\sum_{l=-n/2}^{n/2}\left\{(|z_{i+l,k}| - \overline{|z_{j,k}|}) \times \left(|z'_{j+l,k+l}| - \overline{|z'_{j,k+l}|}\right)\right\},$$

$$|z_{j,k}| = \sqrt{\left(z_s(x_j,t_k)\right)^2 + \left(z_c(x_j,t_k)\right)^2}\ ,$$

$$|z'_{j,k+1}| = \sqrt{\left(z_s(x'_j,t_{k+1})\right)^2 + \left(z_c(x'_j,t_{k+1})\right)^2}\ ,$$

$$\overline{|z_{j,k}|} = \frac{1}{n}\sum_{l=-n/2}^{n/2}|z_{j+l,k}|\ ,$$

where n specifies the range of correlation calculation. The tracking point moves from x to $x' = x + v\Delta T$.

The correlation $\gamma(x_j, x'_j, t_k)$ is high (low), if the velocity derivation error is small (large). Thus the weighted mean velocity $v'(x, t)$ is less affected by the error, and the accuracy of the motion tracking is improved.

C. DP tracking method

In this research, we formulate local myocardial motion on a M-mode ultrasonic Doppler signal as the followings. At first, we define a M-mode ultrasonic Doppler signal which is acquired from a subject as,

$$\boldsymbol{Z} = \{z(x_i,t_k)\}\ , \tag{7}$$

$i = 0, 1, 2, \ldots, M,\ \ k = 0, 1, 2, \ldots, L,$

where *M, L* denotes the number of sampling point along ultrasonic beam and the number of ultrasonic beam contained in one data acquisition, respectively. To simplify the following notations, a set of sampling point on the time t_k is denotes by,

$$\boldsymbol{x}_k = \{x_i|t_k\}\ . \tag{8}$$

Of course, $\boldsymbol{x}_k = \boldsymbol{x}_{k+1}$ for all k.

Next, a tracking start point is located on the depth which is corresponds to the location of myocardium at the tracking start time t_0. We express the tracking point d_0 is equivalent to that of one of the sampling points x_0.

$$d_0 = x_i \mid x_i \in x_0\ . \tag{9}$$

Similarly, a tracking pints at the timing t_k is denoted by,

$$d_k = x_i \mid x_i \in x_k\ . \tag{10}$$

Using above notations, the motion tracking process of which the start point is d_0 is equivalent to determining a set of tracking points,

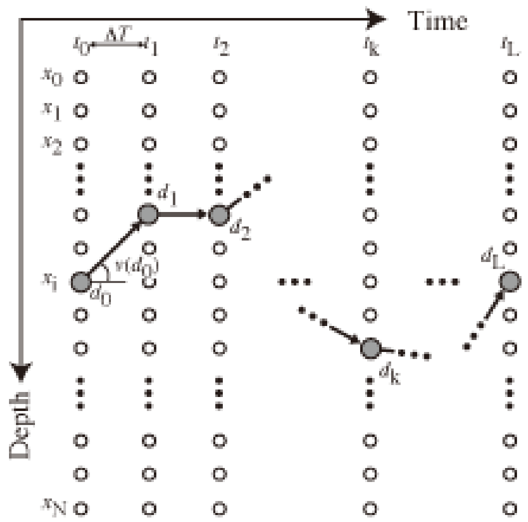


Fig.5 Basic concept of DP tracking method

$$\boldsymbol{D} = (d_0, d_1, d_2, \ldots, d_L)\ , \tag{11}$$

$$d_k = x_i \mid x_i \in x_k\ . \tag{12}$$

When multiple tracking points are employed in tracking, tracking points at t_k are expressed as a multiplet d_k,

$$d_k = \left(d_k^{(0)}, d_k^{(1)}, d_k^{(2)}, \ldots, d_k^{(N)}\right), \tag{13}$$

where N denotes the number of tracking points.

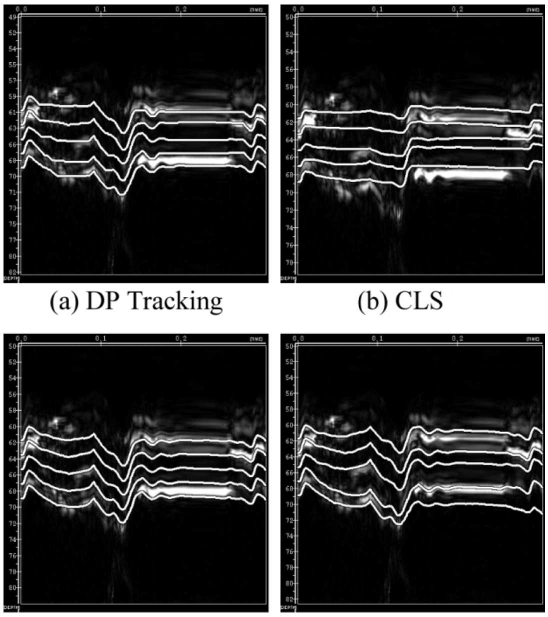
We consider that the motion tracking process is equivalent to determining a trajectory $\boldsymbol{D}$ which minimizes the objective function,

$$f(D) = \sum_{l=1}^{L}\{d_{l-1} + \Delta T v(d_l,t_l) - d_l\}^2 \tag{14}$$

under the assumption of that the start and end points d_0, d_L are equal to each other. The trajectory $\boldsymbol{D}$ which minimizes (14) can be determined using the dynamic programming algorithm. Fig.5 illustrates the basic concept of this motion tracking process.

III. Experiment

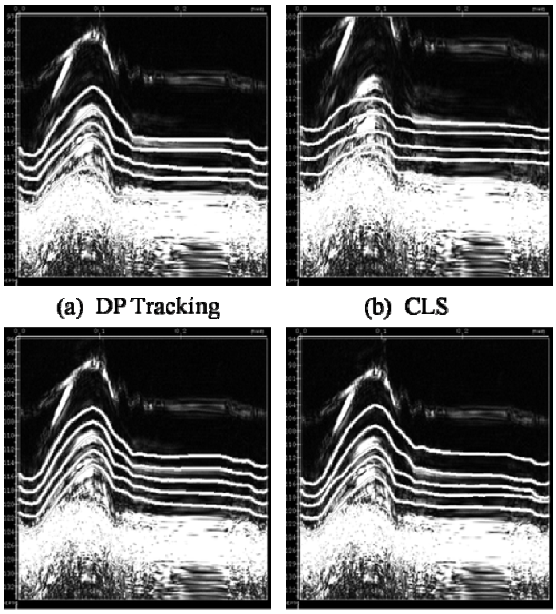
To confirm the tracking performance of the proposed DP tracking method, we conduct an experiment where a clinical ultrasonic signal acquired from a normal subject is used for motion tracking. Visual comparison of tracking results derived using three conventional tracking methods; i.e. the simple tracking, constraint least square tracking method proposed by Kanai [6] and elastic tracking model [10], and the proposed method are shown by Fig.6 and Fig.7. In the figure, upper and lower images correspond to interventricular septum (IVS) and left ventricular posterior wall (LPW), respectively.



(a) DP Tracking (b) CLS

(c) CWP+E (d) Simple Tracking

Fig.6 Visual comparison of tracking results of IVS



(a) DP Tracking **(b) CLS**

(c) CWP + E **(d) Simple Tracking**

Fig.7 Visual comparison of tracking results of LPW.

From the visual comparison, it is observed that the trajectories derived by the three conventional methods contain several amount of tracking error where for instance the start point and end point are significantly different to each other. It is also observed that the DP tracking method (a) takes less error.

IV. Conclusions

We proposed a new method to estimate heart motion from M-mode echocardiograms that uses a dynamic programing method that is well adapted to typical heart dynamics. The DP algorithm is easily realized and brings us more accuracy result in order to help the doctor to analyse the heart disease

Our future work includes quantitative evaluation of the proposed method for motion tracking performance using enough number of clinical ultrasonic data.

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