

Master thesis

Motion Tracking of Local Myocardial Tissue Using DP Tracking Method on M-mode Echocardiogram



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Abstract

Today, cardiovascular diseases have been the second cause of Japanese's death, which result in about 160,000 persons' death every year. Moreover, the diseases of strokes in circulatory organ result in about 30 percents of the total. Due to heart's constancy of pulsation, it is necessary to use high temporal resolution equipments to diagnose the diseases. In the early stages of cardiovascular diseases, symptom is shown in interior of the myocardium, and the entirety of myocardium tends to be enlarged with the progress of symptom. The process of motion, includes two kinds of motion which include pulsation and the motion of displacement within myocardium layer. Therefore, separation of these two motions is required for evaluation of the functions of the local myocardium. Ultrasonic diagnostic equipment is widely used to diagnose fast moving heart on high temporal resolution. By using ultrasonic diagnostic equipment in the performance diagnostics of the heart, the analysis mainly relies on visual inspection or manual measurements by experienced doctors from echocardiogram which is displayed on display. These 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, and automated computer-based analysis is highly desirable to obtain more objective and quantitative diagnoses.

The purpose of this research is to do motion tracking on left ventricular cardiac muscle automatically using echocardiogram with high temporal resolution. M-mode echocardiogram is widely used in clinics to measure diagnostic indexes like thickening and thinning of myocardial muscle layers.

Several approaches have been proposed to quantify the motion of myocardium from two dimensional (2-D) echocardiograms [1, 2, 3, 4]. However, in these approaches, the spacial and temporal resolutions of quantification and tracking of myocardial motion are strongly depends on those of imaging performance of ultrasonic equipment. Particularly, echocardiograms employ low-pass filters on ultrasonic signals in the image generation process. Consequently, the spatial resolution of images becomes low and insufficient for detailed diagnosis of local, inner

myocardial tissues. To enable quantitative evaluation of tissue's kinetic performance on inner myocardial, accurate and suitable tracking of myocardium in high spatial resolution is required. To overcome this low temporal resolution problem, several approaches employ raw ultrasonic signal from which echocardiograms are generated[5, 6, 7, 8, 9, 10].

In this research, 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.

We employ two experiments to estimate the motion tracking using proposed method and three previous methods. One is the visual comparison, the other one is the quantitative evaluation. Either from visual comparison or quantitative evaluation, it clearly observed that the average error of proposed method is the least one in the results of four methods. In this thesis, Chapter 1 describes my research background and objective, and Chapter 2 starts the related technology of present study. The proposed method is described in Chapter 3 in detail. Chapter 4 describes experiments and quantitative evaluation. In the last Chapter, a conclusion is described.

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Chapter 1

Introduction

1.1 Background

Today, cardiovascular diseases have been the second cause of Japanese's death, which result in about 160,000 persons' death every year. Moreover, the diseases of strokes in circulatory organ result in about 30 percents of the total. Medical imaging treatment is widely used to diagnose cardiovascular diseases. Some present advanced techniques are employed to estimate performance of cardiac muscle with image diagnosis, which include MRI (Magnetic Resonance Imaging), X-CT, scintigraphy and echocardiography. Compared with other organs', because of heart beating, image-making of heart is more difficult to be realized. To evaluate the performance of motion, using dynamic image is necessary. Moreover, the equipment with high spatial resolution serves doctors with ability on estimating displacement cardiac muscle layer.

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. On the other hand, in clinical cardiology, the analysis mainly relies on visual inspection or manual measurements by experienced cardiologists. Manual protocols 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.

The 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, 2, 3, 4]. However, in these approaches, the spacial and temporal resolutions of quantification and tracking of myocardial motion are strongly depends on those of imaging performance of ultrasonic equipment. Particularly, echocardiograms employ low-pass filters on ultrasonic signals in the image generation process. The spacial resolution of images consequently becomes low and insufficient for detailed diagnosis of local, inner myocardial tissues. To enable quantitative evaluation for kinetic performance on inner myocardial tissues, accurate and suitable tracking of myocardium in high spacial resolution is required. To overcome this low spatial resolution problem, some motion tracking approaches employ raw ultrasonic signal from which echocardiograms are generated[5, 6, 7, 8, 9, 10]

1.2 Objective of Research

Ultrasonic signal backscattered from myocardium contains several amounts of speckles and noise due to scattering interference of sound wave in muscle layers. These speckles and noise affect the image quality of echocardiogram and sometimes corrupt the assessments for diagnosis. The tracking method is required robustness against such speckles and noise.

In this paper, we propose a novel approach for local, inner myocardial motion tracking.

The proposed approach employs M-mode ultrasonic Doppler signals which are obtained with high spatial resolution to overcome the low resolution problem and Dynamic Programming (DP) based motion tracking method for the robustness against speckles and noise.

DP is a well-known method of solving problems exhibiting the properties of overlapping sub-problems and optimal substructure. In the method, DP is employed to find a trajectory which minimizes an objective function under the periodic assumption on myocardial motion.

Chapter 2

Knowledge and technologies

Ultrasonic diagnosis is a kind of equipment that can obtain information of inner to the body using backscattered ultrasonic which is received when an ultrasonic transducer placed on the chest transmits pulsed ultrasonic toward the heart. The diagnosis which is on basis of ultrasonic serves clinicians and doctors with multifold advantages noninvasiveness of diagnosis, less side effects compared with X rays and γ rays and more effective diagnosis.

Now, ultrasonic diagnosis mainly bases on diagnosis of form of organism's parenchyma using pulsed ultrasonic and evaluation of organism's performance using of blood flow rate by Doppler effect which is obtained from akaryocyte.

2.1 Ultrasonic diagnosis

2.1.1 Property of Ultrasonic

Ultrasonic is cyclic sound pressure with a frequency greater than the upper limit of human hearing. Although this limit varies from person to person, it is approximately 20 kilohertz (20,000 hertz) in healthy, young adults and thus, 20 kHz serves as a useful lower limit in describing ultrasound. The production of ultrasound is used in many different fields, typically to penetrate a medium and measure the reflection signature or supply focused energy. The reflection signature can reveal details about the inner structure of the medium, a property also used by animals such as bats for hunting. One of the most well known application of ultrasonic is its foetus in echocardiography to produce pictures of features in the human ventricle. There are a vast number of other applications as well. Fig.2.7 shows approximate frequency ranges corresponding to ultrasound with rough guide of some applications.

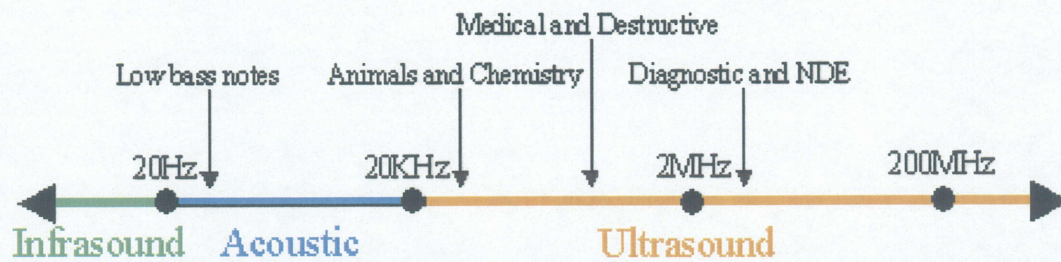


Fig. 2.1: Approximate Frequency ranges corresponding to ultrasound with rough guide of some applications

Medium	Velocity of propagation [m/sec]	Impedance [$10^6 \text{ kg/m}^2 \cdot \text{sec}$]	Attenuate [dB/cm ⁻¹]
Water	1483	1.48	0.0022
Air	343	0.000415	12
Blood	1570	1.61	0.18
Parenchyma	1570	1.63	0.81
Muscle	1585	1.70	1.3~3.3
Bone	4080	7.80	13

Table. 2.1: Acoustic characters of tissues

The acoustic impedance Z is defined by

$$Z = \rho c \quad (2.1)$$

where ρ is density of tissues and c the velocity of propagation. And we can obtain specular reflection R which is defined by,

$$R = \frac{(Z_1 - Z_2)^2}{(Z_1 + Z_2)^2} \quad (2.2)$$

It specifies that more difference between two Z brings larger reflection R . Therefore, it is necessary to avoid using parts of which difference is large, i.e. bone and lung. We show the acoustic characters of different tissues in the body in diagnosis of cardiovascular diseases on Table.2.1.

Some other acoustic characters are expressed as follows,

1. Attenuation of ultrasonic wave which occurs in gas is large, and ultrasonic wave is transmitted well in liquid and solid .

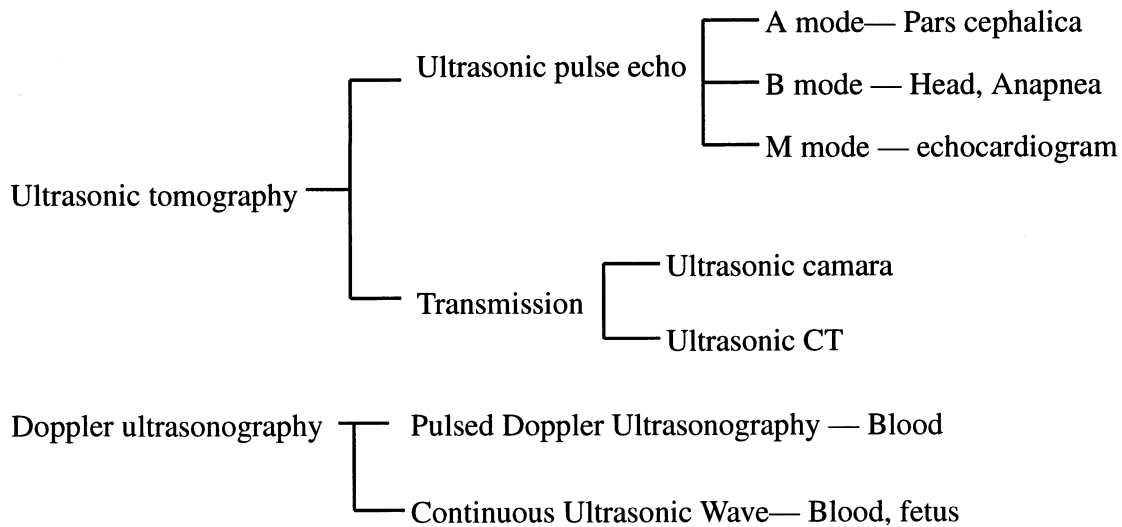


Fig. 2.2: The categories of ultrasonic diagnoses

2. Transmission in liquid and gas occurs as transverse wave, and it can occur in solid not only as transverse wave but also longitudinal wave.
3. The motion of the these particles is cause by 2 factors: the pressure of the wave (which forces them to move in the beginning) and the forces of the restoring molecules (also known as the elasticity of the medium)
4. The amount of particle movement is dependent on the pressure change associated with the wave, therefore the increased pressure change equals the increased particle movement, and therefore louder sound.

2.1.2 The categories of ultrasonic diagnoses

It is divided into Doppler Ultrasonic and ultrasonography according to the principle of ultrasonic diagnoses. Fig.2.2 shows the categories of ultrasonic diagnoses and main applictions.

2.1.3 Ultrasonic echo

In this research, we employ ultrasonic pulse echo technique to conduct analysis. Fig.2.3 illustrates the principle of pulse-echo technique. The top part of the figure shows the physical

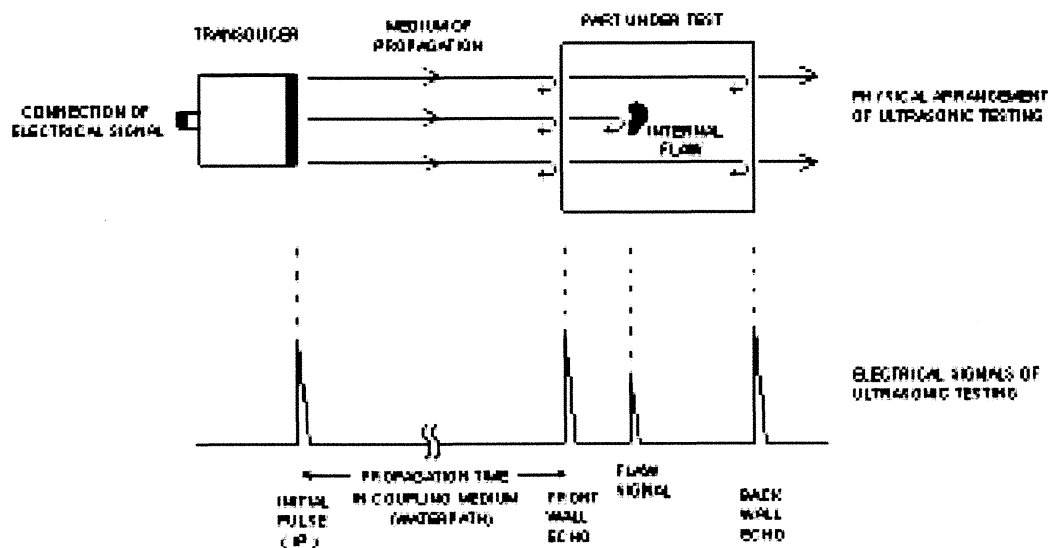


Fig. 2.3: Basic principle of ultrasonic testing by pulse-echo method.

configuration of the test. The transducer produces a pressure wave in response to the electrical pulse which has been applied to it. This is usually referred to as an ultrasonic pulse or initial pulse. The pressure wave travels through the coupling medium, which is usually water, to the test part. At the interface of the coupling medium and the test part the ultrasonic pulse enters the part but a portion is reflected back to the transducer. After entering the part there is a partial reflection from the internal flaw. The remainder of the pulse travels to the back wall, where another partial reflection and transmission occurs. The lower part of Fig.2.3 shows the electrical signals as a function of time, referenced to the locations of the transducer and the test part with an internal flaw.

Three kinds of ultrasonic pulse echo techniques have been employed generally in recent years, **A mode** When the ultrasonic beam encounters an anatomic boundary, the received sound impulse is processed to appear as a vertical reflection of a point. On the display, it looks like spikes of different heights (the amplitude). The intensity of the returning impulse determined the height of the vertical reflection and the time it took for the impulse to make the round trip would determine the space between verticals. The distance between these spikes can be measured accurately by dividing the speed of sound in tissue (1540 m/sec) by half the sound travel time.

B mode Also called B-mode echography, B-mode sonography, 2D-mode, and sonogram. B-mode ultrasound (Brightness-mode) is the display of a 2D-map of B-mode data, currently the most common form of ultrasound imaging. The development from A-mode to B-mode is that the

ultrasound signal is used to produce various points whose brightness depends on the amplitude instead of the spiking vertical movements in the A-mode. Sweeping a narrow ultrasound beam through the area being examined while transmitting pulses and detecting echoes along closely spaced scan lines produces B-scan images. The vertical position of each bright dot is determined by the time delay from pulse transmission to return of the echo, and the horizontal position by the location of the receiving transducer element. To generate a rapid series of individual 2D images that show motion, the ultrasound beam is swept repeatedly. The returning sound pulses in B-mode have different shades of darkness depending on their intensities. The varying shades of gray reflect variations in the texture of internal organs. This form of display (solid areas appear white and fluid areas appear black) is also called gray scale.

M mode The M-mode (Motion-mode) ultrasound is used for analyzing moving body parts (also called time-motion or TM-mode) commonly in cardiac and fetal cardiac imaging. The application of B-mode and a strip chart recorder allows visualization of the structures as a function of depth and time. The M-mode ultrasound transducer beam is stationary while the echoes from a moving reflector are received at varying times. A single beam in an ultrasound scan is used to produce the one-dimensional M-mode picture, where movement of a structure such as a heart valve can be depicted in a wave-like manner. The high sampling frequency (up to 1000 pulses per second) is useful in assessing rates and motion, particularly in cardiac structures such as the various valves and the chamber walls.

In my research, A mode is seldom employed in clinical analysis because of some drawbacks that we don't know what the object generating the echo looks like and also don't know for sure what the echo bounced off of. B mode and M mode are widely used in clinics at the present day. In this thesis, M mode echocardiogram is used to diagnose and treat the cardiovascular diseases as an essential and necessary tool.

2.1.4 Scan method of ultrasonic transducer

Ultrasonic transducer arrays are broadly classified under the categories of linear scan type and sector scan type. Conventional linear scan type arrays comprise piezoelectric transducers, 256 in number, which are successively linearly arranged side by side. A group of 16 transducers is selectively activated by delayed burst pulses generated by a commonly shared transmit circuitry so that a focused ultrasonic beam is transmitted. The selected group is successively shifted to the next by one transducer element to shift the beam linearly to the next, so that the ultrasonic energy is scanned in a rectangular format. The linear scan system has a disadvantage in that it is incapable of scanning areas behind ribs and in that the transducer array is relatively bulky for

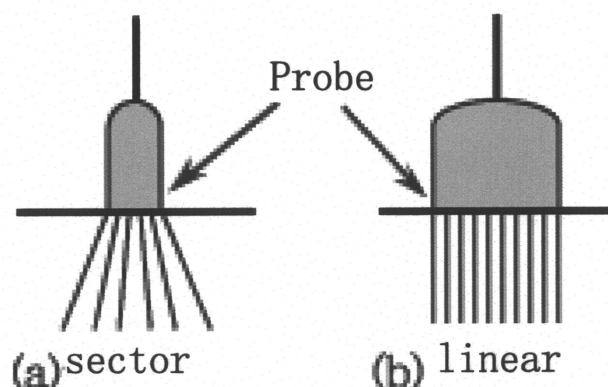


Fig. 2.4: Scan method of ultrasonic beam

manipulation.

Conventional transducer arrays of the sector scan type, on the other hand, usually comprise 32 transducer elements each of which is associated with its own transmit and receive circuitry. The transmit circuitry steers the ultrasonic beam in a sector format by applying successively delayed burst pulses to the transducers. Therefore, in diagnoses of cardiovascular diseases, we must use sector scan to read information from intercostal. Fig.2.4 shows the difference between two methods.

2.1.5 The spatial resolution and temporal resolution

M-mode Doppler is a pulsed Doppler technique that allows to obtain a spatio-temporal velocity map with a temporal resolution of 5 ms, a spatial resolution of $300\ \mu\text{m}$, and a velocity resolution of about 3 cm per sec. A major difference exists between M-mode Doppler and normal pulsed Doppler as the latter yields information on the temporal course of velocity at a fixed spatial point while color M-mode Doppler allows acquisition of information on velocity, time and space. By obtaining information on these three parameters, a more complete picture of the filling pattern can be acquired.

Fig.2.5 shows a basic principle that we can obtain M-mode echocardiogram along spatial resolution and temporal resolution from backscattered ultrasonic of pulsed ultrasonic toward the heart which is transmitted by a ultrasonic transducer placed on the chest.

2.1.6 Disadvantages and advantages of ultrasonic diagnose

Strengths

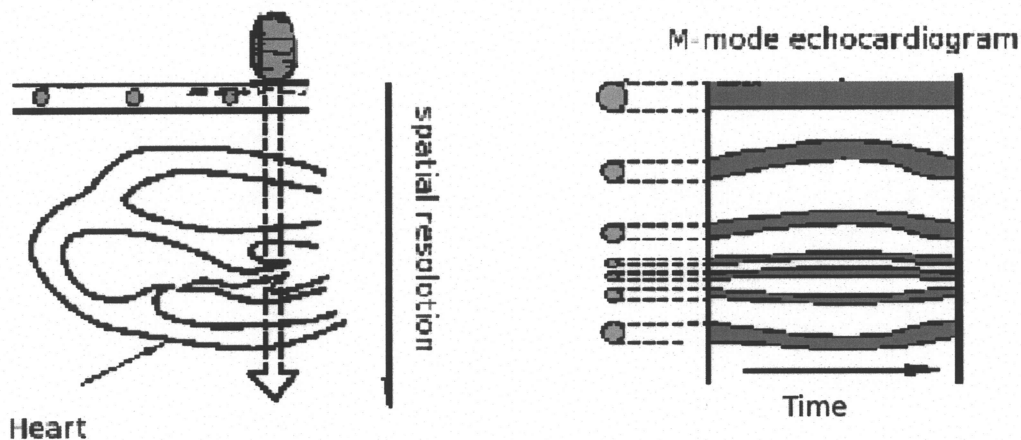


Fig. 2.5: Spatial resolution and temporal resolution

1. It images muscle, soft tissue, and bone surfaces very well and is particularly useful for delineating the interfaces between solid and fluid-filled spaces.
2. It renders "live" images, where the operator can dynamically select the most useful section for diagnosing and documenting changes, often enabling rapid diagnoses. Live images also allow for ultrasound-guided biopsies or injections, which can be cumbersome with other imaging modalities.
3. It shows the structure of organs.
4. It has no known long-term side effects and rarely causes any discomfort to the patient.
5. Equipment is widely available and comparatively flexible.
6. Small, easily carried scanners are available; examinations can be performed at the bedside.
7. Relatively inexpensive compared to other modes of investigation, such as computed X-ray tomography, DEXA or magnetic resonance imaging.
8. Spatial resolution is better in high frequency ultrasound transducers than it is in most other imaging modalities.

Weaknesses

1. Sonographic devices have trouble penetrating bone. For example, sonography of the adult brain is very limited though improvements are being made in transcranial ultrasonography.
2. Sonography performs very poorly when there is a gas between the transducer and the organ of interest, due to the extreme differences in acoustic impedance. For example,

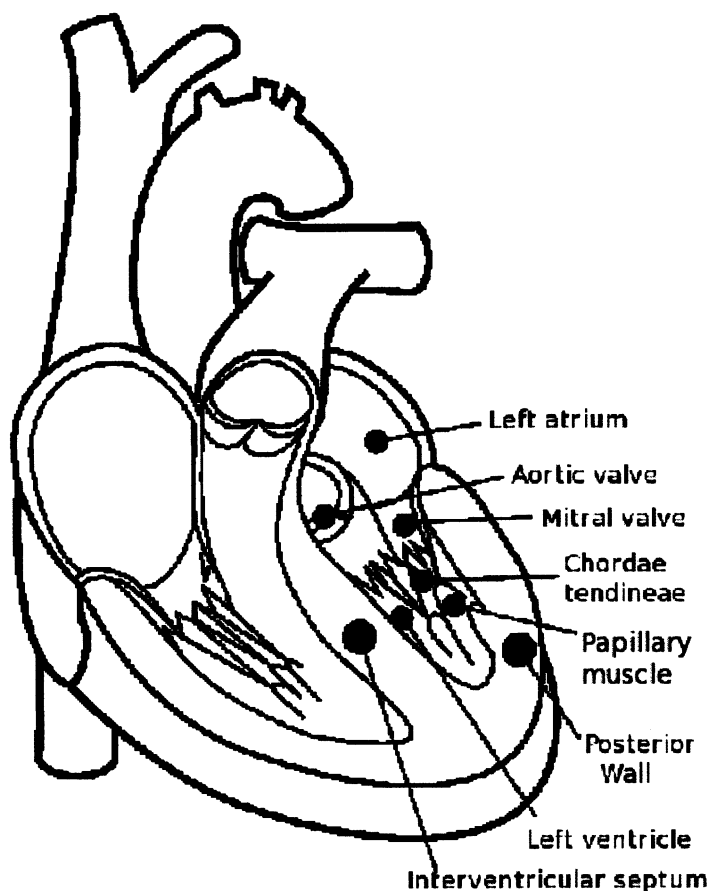


Fig. 2.6: The structure of heart

overlying gas in the gastrointestinal tract often makes ultrasound scanning of the pancreas difficult, and lung imaging is not possible (apart from demarcating pleural effusions).

3. Even in the absence of bone or air, the depth penetration of ultrasound may be limited depending on the frequency of imaging. Consequently, there might be difficulties imaging structures deep in the body, especially in obese patients.
4. The method is operator-dependent. A high level of skill and experience is needed to acquire good-quality images and make accurate diagnoses.
5. There is no scout image as there is with CT and MRI. Once an image has been acquired there is no exact way to tell which part of the body was imaged.

Risks and side-effects

Ultrasonography is generally considered a "safe" imaging modality. However slight detrimental effects have been occasionally observed (see below). Diagnostic ultrasound studies of the foetus are generally considered to be safe during pregnancy. This diagnostic procedure should

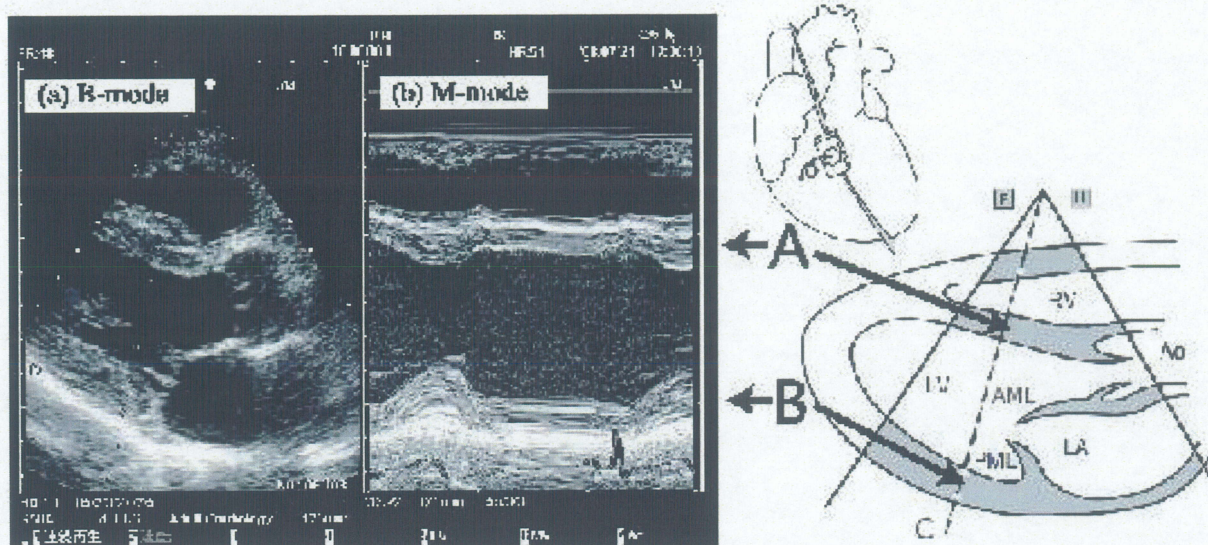


Fig. 2.7: Example of clinical echocardiogram: (a) B-mode, (b) M-mode echocardiogram, A. Interventricular septum and B. Left ventricular posterior wall

be performed only when there is a valid medical indication, and the lowest possible ultrasonic exposure setting should be used to gain the necessary diagnostic information under the "as low as reasonably achievable" or ALARA principle.

World Health Organizations technical report series 875(1998). Supports that ultrasound is harmless: "Diagnostic ultrasound is recognized as a safe, effective, and highly flexible imaging modality capable of providing clinically relevant information about most parts of the body in a rapid and cost-effective fashion". Although there is no evidence ultrasound could be harmful for the foetus, US Food and Drug Administration views promotion, selling, or leasing of ultrasound equipment for making "keepsake foetal videos" to be an unapproved use of a medical device.

2.2 The structure of Heart

The weight of the average human heart is 200–300g. Heart is made up of four chambers. The right atrium and ventricle help to circulate blood through the lungs. The left atrium and ventricle help to circulate blood to the rest of your body.

Between these four chambers there are valves. These valves help to keep blood flowing in the right direction. They also prevent it from slipping back to the wrong chamber when the heart is at rest. In a sense, they are one-way doors.

Fig.2.6 shows a structure of heart and describes the principle parts by arrows.

Fig.2.7 shows an example of clinical echocardiogram which is used in our research. In the

figure, (a) and (b) are B-mode and M-mode echocardiograms, respectively. We can obtain (b) from (a) along the scan broken line C.

Chapter 3

The proposed method

In this paper, we propose a novel approach for local, inner myocardial motion tracking. The proposed approach employs M-mode ultrasonic Doppler signals which are obtained with high spatial resolution to overcome the low resolution problem and Dynamic Programming (DP) based motion tracking method for the robustness against speckles and noise.

3.1 Dynamic Programming

Dynamic Programming is a well-known method of solving problems exhibiting the properties of overlapping sub-problems and optimal substructure. It is both a mathematical optimization method, and a computer programming method. In both contexts, it refers to simplifying a complicated problem by breaking it down into simpler sub-problems in a recursive manner. While some decision problems cannot be taken apart this way, decisions that span several points in time do often break apart recursively. Bellman called this the "Principle of Optimality". Likewise, in computer science, a problem which can be broken down recursively is said to have optimal sub-structure.

If sub-problems can be nested recursively inside larger problems, so that dynamic programming methods are applicable, then there is a relation between the value of the larger problem and the values of the sub-problems. In the optimization literature this relationship is called the Bellman equation.

Fig.3.1 shows a process that finding the shortest path in a graph using optimal substructure by Dynamic Programming. In figure, we can find that the a wavy line indicates a shortest path between the two vertices it connects (other nodes on these paths are not shown) and the bold line is the overall shortest path from start to goal. In my research, DP is employed to find a trajectory which minimizes an objective function under the periodic assumption on myocardial motion.

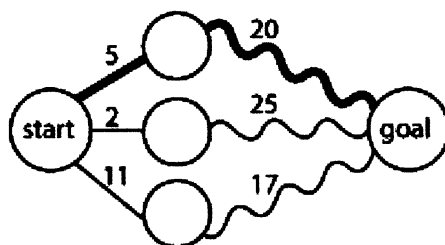


Fig. 3.1: A application of Dynamic Programming

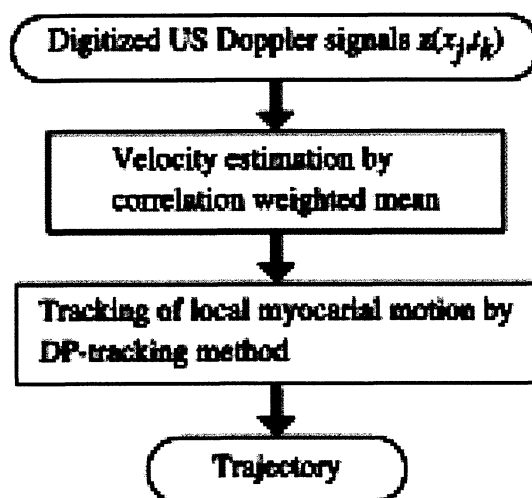


Fig. 3.2: The flow of motion tracking

3.2 The proposed method

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.3.2. 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 determines the optimal trajectory of the tracking points.

3.2.1 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

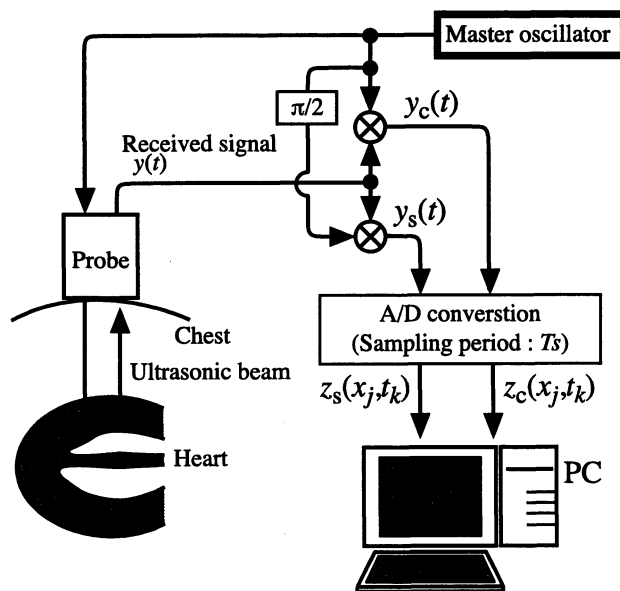


Fig. 3.3: Block-diagram of data acquisition system

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.3.3 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 backscattered 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 \mathbf{Z} . Discrete Doppler signal $\mathbf{Z} = \{z(x_i, t_k)\}$ 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,

$$\begin{aligned} z(x_i, t_k) &= z_s(x_i, t_k) + jz_c(x_i, t_k), \\ x_i &= ic_0T_s, \\ t_k &= k\Delta T, \end{aligned} \quad (3.1)$$

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

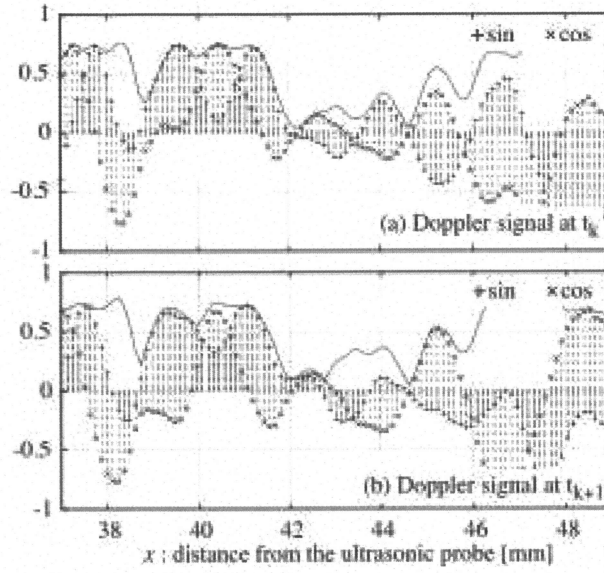


Fig. 3.4: Example of acquired ultrasonic Doppler signal

3.2.2 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].

$$\theta(x_j, t_k) = \tan^{-1} \frac{z_c(x_j, t_k)}{z_s(x_j, t_k)}, \quad (3.2)$$

$$\Delta\theta(x_j, t_k) = \theta(x_j, t_{k+1}) - \theta(x_j, t_k), \quad (3.3)$$

$$v(x_j, t_k) = \frac{c_0 \Delta\theta(x_j, t_k)}{4\pi f_0 \Delta T}. \quad (3.4)$$

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 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

$$v'(x_j, t_k) = \frac{\sum_{l=0}^{N-1} \gamma(x_{j+l}, x'_{j+l}, t_k) v(x_{j+l}, t_k)}{\sum_{l=0}^{N-1} \gamma(x_{j+l}, x'_{j+l}, t_k)}, \quad (3.5)$$

where N is length (sample size) of ultrasonic pulse, and $\gamma(x_j, x'_j, t_k)$ is the correlation of the magnitude around the sampling point before and after a step of motion tracking, which is defined by,

$$\gamma(x_j, x'_j, t_k) = \frac{|z_{j,k}| \bullet |z'_{j,k+1}|}{\sqrt{|z_{j,k}| \bullet |z_{j,k}|} \sqrt{|z'_{j,k+1}| \bullet |z'_{j,k+1}|}}, \quad (3.6)$$

$$\begin{aligned} & |z_{j,k}| \bullet |z'_{j,k+1}| \\ &= \sum_{l=-n/2}^{n/2} \left\{ (|z_{j+l,k}| - \overline{|z_{j,k}|}) \times (|z'_{j+l,k+1}| - \overline{|z'_{j,k+1}|}) \right\}, \end{aligned}$$

$$\begin{aligned} |z_{j,k}| &= \sqrt{(z_s(x_j, t_k))^2 + (z_c(x_j, t_k))^2}, \\ |z'_{j,k+1}| &= \sqrt{(z_s(x'_j, t_{k+1}))^2 + (z_c(x'_j, t_{k+1}))^2}, \\ \overline{|z_{j,k}|} &= \frac{1}{n} \sum_{l=-n/2}^{n/2} |z_{j+l,k}|, \end{aligned}$$

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

The correlation $\gamma(x, x', t)$ 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.

3.2.3 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,

$$\begin{aligned} Z &= \{z(x_i, t_k)\}, \\ i &= 0, 1, 2, \dots, M, \quad k = 0, 1, 2, \dots, L, \end{aligned} \quad (3.7)$$

where M, L denotes the numbers of sampling point along a ultrasonic beam and 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 $x_k = \{x_i | t_k\}$, $x_k = x_{k+1}$ for all k . Next, a tracking start point is located on the depth which is corresponds to the location of myocardium at the

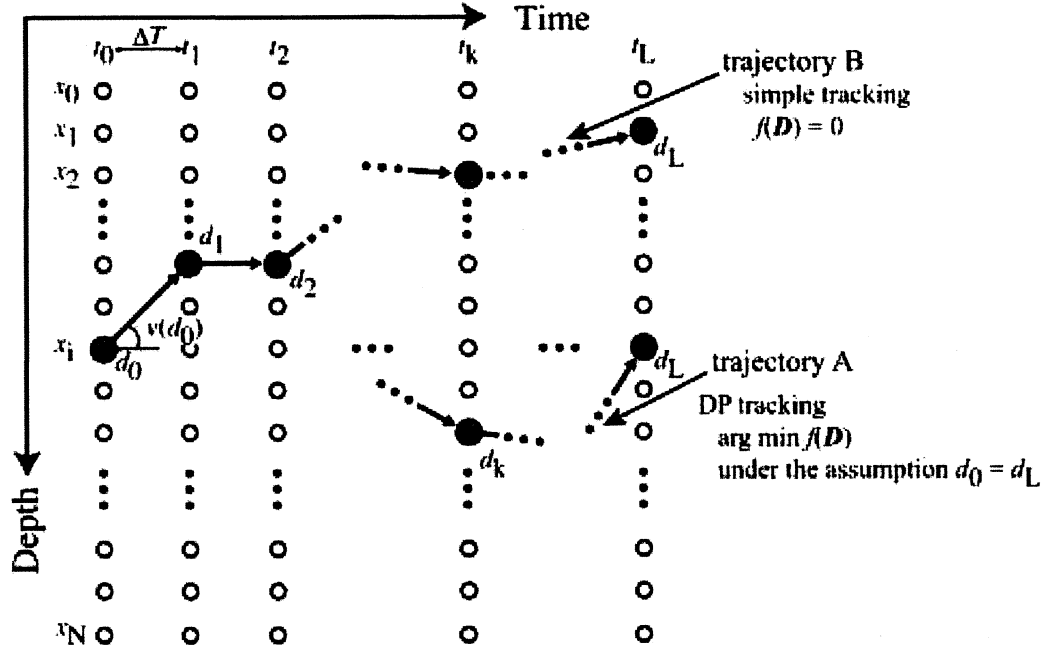


Fig. 3.5: Basic concept of DP tracking method

tracking start time t_0 . We express the tracking point itself in the depth d_0 . The depth of tracking start point d_0 is identical to that of one of the sampling points x_0 :

$$d_0 = x_i \mid x_i \in x_0. \quad (3.8)$$

Similarly, a tracking point at the time t_k is denoted by:

$$d_k = x_i \mid x_i \in x_k. \quad (3.9)$$

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

$$D = (d_0, d_1, d_2, \dots, d_k, \dots, d_L), \quad (3.10)$$

$$d_k = x_i \mid x_i \in x_k. \quad (3.11)$$

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

$$d_k = (d_k^{(0)}, d_k^{(1)}, d_k^{(2)}, \dots, d_k^{(\mathcal{N})}), \quad (3.12)$$

where \mathcal{N} denotes the number of tracking points.

We formalize the motion tracking process as the one of determining a trajectory which minimizes an objective function,

$$f(D) = \sum_{l=1}^L \{d_{l-1} + \Delta T v(d_{l-1}, t_{l-1}) - d_l\}^2, \quad (3.13)$$

under the assumption that the start and end points are the same in the depth, i.e. $d_0 = d_L$. The trajectory D which minimizes (3.13) can be determined using the dynamic programming algorithm.

Fig.3.5 illustrates the basic concept of this motion tracking process. Trajectory A represents a trajectory determined by the DP tracking. As described above, this trajectory is determined as that minimizes the objective function $f()$ under the assumption $d_0 = d_L$. On the other hand, trajectory B denotes a tracking result by the simple tracking algorithm. The simple tracking algorithm simply accumulates instantaneous displacement of a tracking point, $v(d_l, t_l)$, onto the position of the tracking point. Therefore, the objective function $f()$ is zero over one heart-beat tracking period, while it is not guaranteed that the start point d_0 and the end point d_L are the same in the depth. When the boundary condition $d_0 = d_L$ is removed, the DP and the simple tracking methods derive the same trajectory.

It is generally known that the computation time for the DP is $O(n^2)$. n denotes to the number of sampling points contained in searching region at each time t_l . Although this is greater than that of simple tracking, which is constant-order, it is significant decrease from that of the brute-force algorithm.

3.3 Some conventional approaches

Several approaches employ raw ultrasonic signal from which echocardiograms are generated. They are used to compare with the proposed method in the experiment; i.e. the simple tracking, constraint least square tracking method proposed by Kanai[6] and elastic tracking model[10]. In this part, they are simply illustrated as follows, respectively.

3.3.1 Constraint Least-Square method:CLS

The velocity measurement for myocardium can be obtained in the paper[6], and the position d_{k+1} of motion tracking points at $t_{k+1} = t_k + \Delta T$ can be express as,

$$d_{k+1} = d_k + \Delta T v'_k, \quad (3.14)$$

and multiple tracking points are employed in tracking, tracking points at t_k are expressed as a vector (3.12). We assume repetition period of the ultrasonic pulse ΔT as 24 [Point], and n which

specifies the range of correlation calculation in terms of the method which is proposed in the paper[6].

3.3.2 Simple Phase Difference: Simple Tracking

The velocity measurement can be expressed as equation (3.4), equation (3.12) and (3.14) are used to calculate the position of tracking points.

3.3.3 Correlation Weighted Phase-difference with Elastic model:CWP+E

The velocity measurement can be expressed as (3.4) and (3.5). In the process of motion tracking, a elastic modulus $\varepsilon_k=(\varepsilon_{1,k},\varepsilon_{2,k},\varepsilon_{3,k},\dots,\varepsilon_{M,k})^T$ which is a vector of correction is added to equation(3.14) as follow,

$$d_{k+1} = d_k + \Delta T v'_k + \varepsilon_k, \quad (3.15)$$

We consider that the motion tracking process is equivalent to determining a vector which minimizes the objective function,

$$e(\varepsilon_k) = \sum_{j=1}^M \varepsilon_{j,k}^2 + \frac{E}{2} \sum_{j=1}^{M-1} \{(d'_{j+1,k+1} - d'_{j,k+1}) - (d_{j+1} - d_{j,0})\}^2, \quad (3.16)$$

where M is the number of tracking points at t_k and E is a constant which is defined as 0.030 by manual in our experiment. (3.16) is defined as a summation of square of difference of distance between tracking points with constant E.

Chapter 4

Experiment and Result

To evaluate the tracking accuracy of the proposed DP-tracking method, we conduct motion tracking experiments using clinical ultrasonic signals.

M-mode ultrasonic Doppler signals were acquired from three normal subjects using ultrasonic diagnostic equipment (HITACHI EUB-6500 Advanced). The number of acquired ultrasonic signals was 43 of which 16 and 27 target inter-vascular septum (IVS, Figure 2.7 A) and

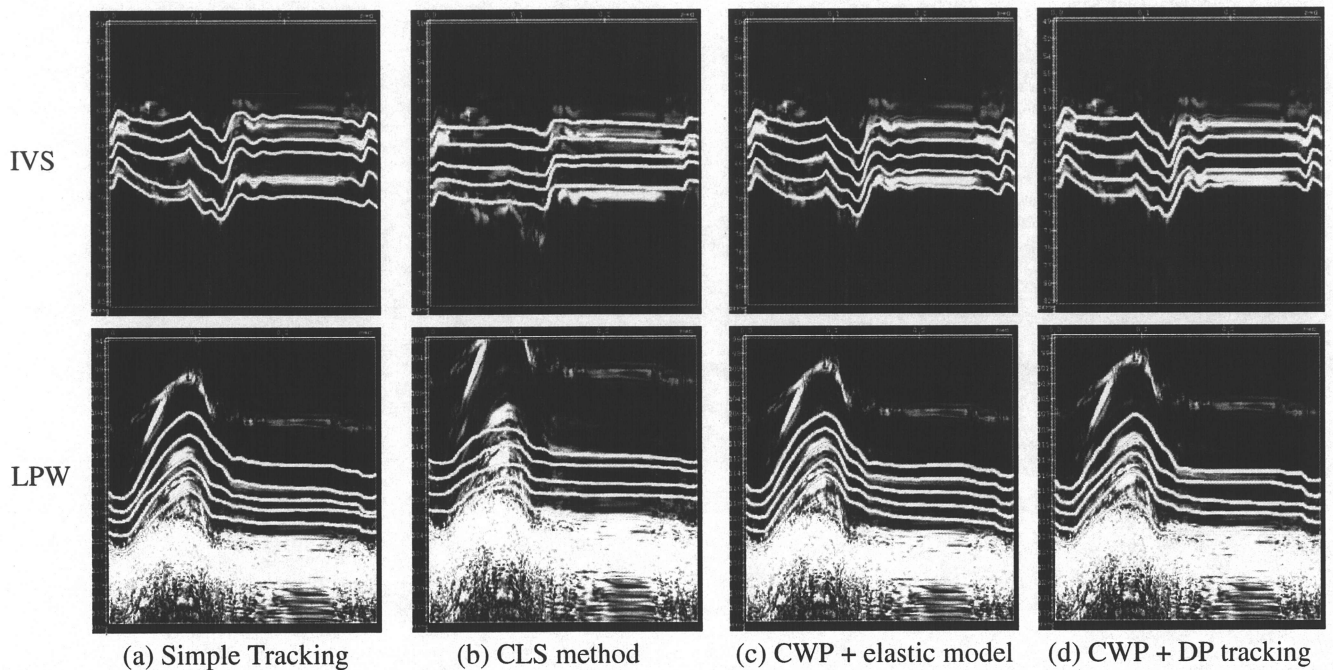


Fig. 4.1: Visual comparison of tracking results: (a) to (d) are tracking results by the Simple Tracking, Constraint Lease Square method proposed by Kanai[6], Elastic tracking model[10] and the proposed DP tracking method, respectively.

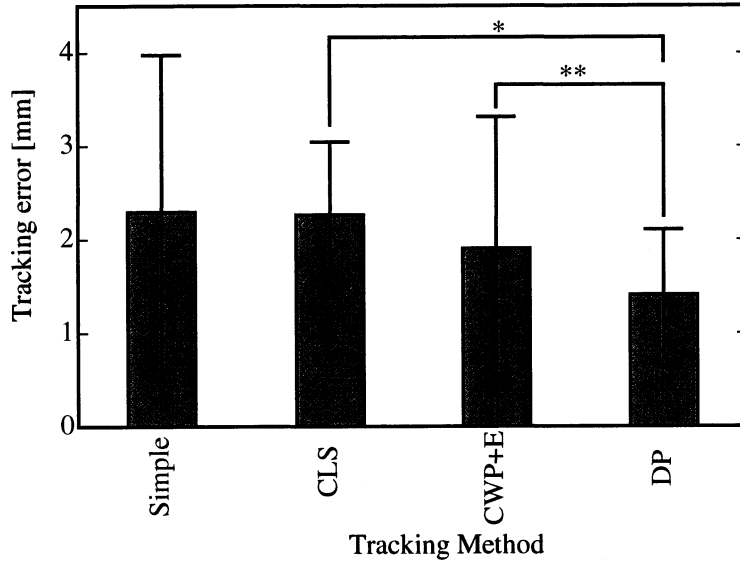


Fig. 4.2: Statistical evaluation of tracking error among four tracking methods.

left-ventricular posterior wall (LVPW, Figure 2.7 B), respectively. The starting points of motion tracking were manually determined on myocardial tissues by trained medical image researcher for each ultrasonic signal.

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 Figure 4.1. In the figure, upper and lower images correspond to IVS and LVPW, respectively. In these results, the number of tracking points is 5. From the visual comparison, it is observed that the trajectories derived by the three conventional methods contain 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 (d) takes less error.

For quantitative evaluation of tracking accuracy of proposed tracking method, we employed the difference of two trajectories obtained by automatic and manual tracking as a measure of tracking accuracy. An operator first traces the motion of endocardium (inside of heart) and epicardium (outside of heart). The four automatic tracking methods are also applied to trace the motion of local myocardium in the same region as manual tracking.

The difference of trajectory E is defined by,

$$E = \frac{\sum_{k=0}^L e(k,k)}{L}, \quad (4.1)$$

$$e(k, k) = \sqrt{\frac{(k-k)(k-k)}{\mathcal{N}}}, \quad (4.2)$$

where $d_k = (d_k^{(0)}, d_k^{(1)}, d_k^{(2)}, \dots, d_k^{(\mathcal{N})})$ and $a_k = (a_k^{(0)}, a_k^{(1)}, a_k^{(2)}, \dots, a_k^{(\mathcal{N})})$ are tracking points at time t_k by automatic and manual, respectively. The number of tracking points \mathcal{N} is 5. L is the number of pulse repetition in one heartbeat.

Figure 4.2 shows the result of evaluation test. The bars and error-bars denote the mean and the standard deviation of difference of trajectory, respectively. The proposed method yielded the minimum mean difference and standard deviation. Statistical significance of the proposed DP-based tracking method against the two conventional method, i.e CLS (indicated by *) and CWP+E (indicated by **) is assessed by paired t -test.

Chapter 5

Conclusion

5.1 Conclusion

We proposed a new method to estimate heart motion from M-mode echocardiograms based on the dynamic programming, which is well adapted to typical heart dynamics. The DP algorithm is easily realized and brings us more accurate result in order to help the doctor to analyze the heart disease. The advantages of the proposed method are summarized as follows.

1. The DP-based tracking method is less affected by the speckle noise, and the accuracy of the motion tracking is improved.
2. The tracking error in the portion with severe attenuation and scatter of ultrasonic signals is suppressed.

5.2 Future Research

Studies on

1. further improvement of the motion tracking accuracy,
2. reducing of the computation time for real-time application, and
3. clinical application of the motion tracking,

are remaining as future research topics.

AppendixA

A.1 Source code of Programming

/home/xserve0/users/chougou/Desktop/Programming/DMMT_new

A.1.1 Source code list

callback.h	Header file of callback function
global.h	global header file of Programming
gui.h	graphic function headfile
manual_tracking.h	manual method headfile
tracking_misc.h	Header file of depicting motion tracking
dmmt_prototypes.h	Header file of declaring tracking result fuctions
dmmt_types.h	headfile of types of motion tracking
dp_tracking.h	Header file of dp functions
main.c	main file of programming
dp_tracking.c	some fuctions of dp programming
tracking_misc_thread.c	main fuction of dp programming
drawing_misc.c	drawing fuctions
manual_tracking.c	functions of manual tracking method
magnitude.c	magnitude functions
select_area.c	functions of selecting area of mycardiogram
manual_tool_window.c	functions of producing windows of manual tool
diff_of_trj.c	functions of calculating difference
average.c	calculating average of difference
README	README

A.1.2 Directory

<code>~/Programming/DMMT_new/src</code>	calculating average of difference
<code>~/Programming/DMMT_new/sample_data</code>	the experiment data
<code>~/Programming/DMMT_new/result</code>	the motion tracking result
<code>~/Programming/DMMT_new/diff_of_traject</code>	the source code of calculating the difference
<code>~/Programming/DMMT_new/manual_ohyama</code>	The manual motion tracking result

A.2 How to run the programming

A.2.1 Motion tracking on windows

```
> setenv XLIB_SKIP_ARGB_VISUALS 1
> make clean
> ./DMMT [OPTIONS] INPUT_DOPPLER_FILE
```

If you want to add some options in your command, it refers to the readme file which is saved in `~ chougou/Desktop/Programming/DMMT_new/src`.

After you run the command, a window appears. You can choose the different kinds of automatic motion tracking methods by options on the window, and then press 'Run' button, an automatic motion tracking result is displayed on the screen. If you want to save the motion tracking result, you can press 'Save' button, and input the name into command window.

A.2.2 Calculate the difference

```
> ./diff_of_traject [input Manual TRAJECTORY file name] [input Automatic TRAJECTORY
file name]
```

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