4D Visualization of the Left Ventricle, Using a Rotating Ultrasound Probe During 4 Cardiac Cycles*

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A new type of external ultrasound probe is proposed to visualize dynamically the left-ventricular three-dimensional deformations. This probe acquires successive apical long axis cross sections by rotating continuously around its axis (one or two rotations per cardiac cycle). From the set of one hundred images obtained during only four consecutive heart beats the volume is reconstructed. In the present development phase, segmentation is performed manually, but supervised in order to guarantee a coherent set of contours. Experimental validations on regular cardiac cycles have been made on a patient fitted with pacemaker. The results obtained are promising.

Keywords: Instrumentation technology, 4D ultrasound imaging, heart imaging, rotating probe.

Introduction

In cardiology, ultrasound imaging is an established alternative to other imaging techniques such as X-ray imaging, magnetic resonance imaging (MRI), or nuclear imaging. It is a fast, non-invasive and low-cost diagnostic technique.

Most heart diseases can be observed on ultrasound images, provided by a fixed two–dimensional (2D) probe, at the video acquisition rate (25 images per second) [Shiller et al. 1989]. Unfortunately, these 2D-images do not provide accurate and reliable measures of positions, shapes and volumes of heart cavities [Fenster et al. (1996)]. The 3D echocardiography allows a better characterization of several major heart functions and the accurate estimation of usual clinical measures such as ejection fraction, global cardiac flow, intra-cardiac volume-pressure ratio, etc [Greenleaf et al. (1993)], [Belohlavek et al. (1993)].

Scanning *in vivo* the LV volume by an ultrasound beam is therefore a matter of research in ultrasound imaging. The scanning is performed in consecutive sectors obtained by moving a 2D probe. Several results have already been reported with various trans–thoracic or trans–oesophageal 2D probes that can move in parallel planes [Matsumoto et al. (1981)], oscillate [Martin et al. (1990)], or rotate [McCann et al. (1988)], [Nosdir et al. (1996)].

The successive images acquired during several heart beats, assumed to be identical, are grouped as a function of their relative instant during the cardiac cycle. After the manual [Gustavson et al. (1993)] or automatic [Coppini et al. (1995)] segmentation of these images, the contours of the endocardial wall are mapped onto a 3D system of co-ordinates. Finally, the volume is obtained by adjusting the parameters of a 3D model of the LV [Moritz et al. (1983)], [McCann et al. (1988)], [Greenleaf et al. (1993)], [Belohlavek et al. (1993)].

In our case, the proposed device is based on the trans–thoracic rotating probe principle (see Fig. 1). This technical choice offers several advantages. First of all, it is used in non-invasive clinical investigations. Secondly, the ultrasound probe does not require expensive and critical

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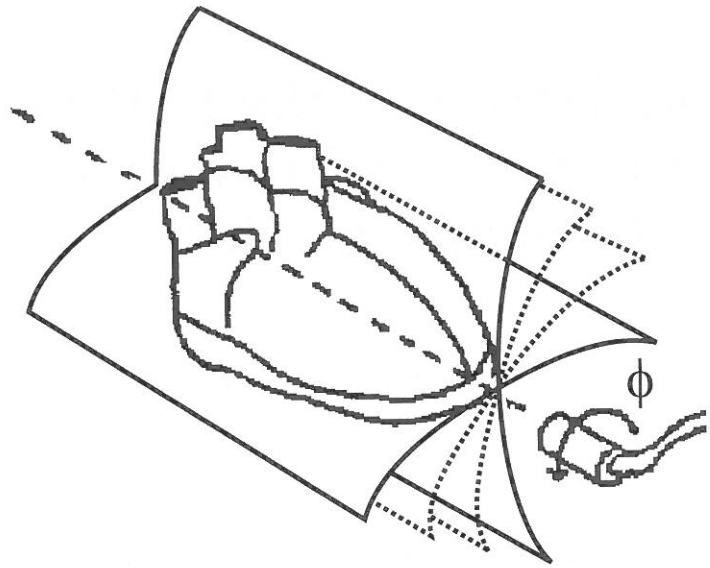

Fig. 1. Principle of the rotating ultrasound probe (ϕ is the rotation angle).

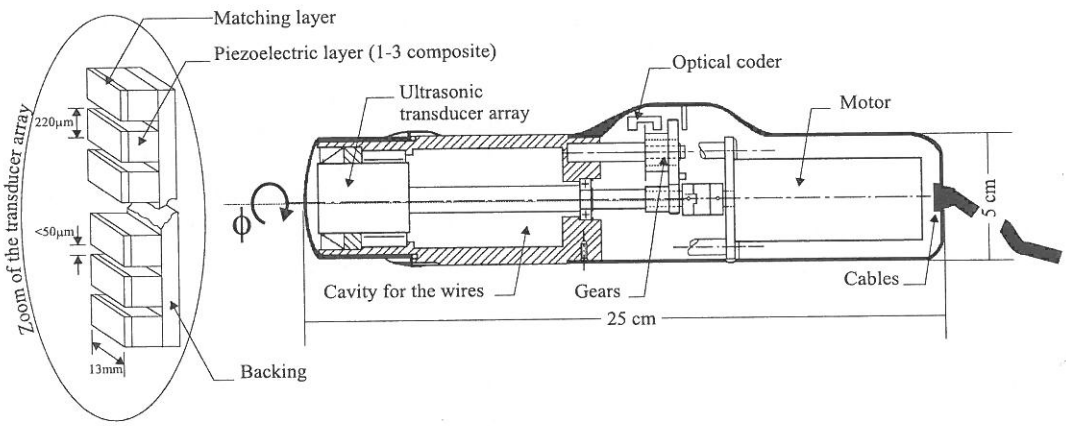

Fig. 2. Schematic view of the rotating probe. ϕ is the rotation angle.

miniaturization technology. Moreover, the periodic movement law of the probe allows the natural application of harmonic models to reconstruct the LV.

This paper presents several results obtained with a new external rotating 2D probe, specially designed for fast 3D ultrasound imaging. Its detailed description is given in the following section. The supervising method used by the expert to trace the contours is then described and the technique used to render the successive LV volumes is explained. The last section presents several clinical results obtained for a typical patient with the heart controlled by a pacemaker.

1. Experimental Device

The 2D rotating probe (see Fig. 2) has cylindrical symmetry with respect to the rotation axis. It is composed of two parts : the motor and the ultrasound sensor. The probe is controlled by a standard ultrasound system through specific electronic boards. An electrocardiogram (ECG) acquisition system can be used to synchronize the acquisition.

1.1. Motor

The electrical motor makes the ultrasound sensor rotate. An optical coder is used to measure the rotation angle, ϕ, of the sensor each 7.11 ms.

The motor runs continuously at a speed of up to 500 revolutions per minute, but in the experiments detailed in this paper, the operating speed is set to 200 revolutions per minute. Due to the twisting of the sensor liaison wires, the motor speed must be inverted after a maximum of 2 rotations. The inversion time is very short (< 2.5 ms) and may be neglected.

1.2. Ultrasound Sensor

The ultrasound sensor is perpendicular to the rotation axis. It is made up of 64 piezoelectric elements. Their dimensions are given in Fig. 2. The focal distance is about 10 cm from the sensor. The frequency bandwidth at 6 dB is limited by the two cut-off frequencies 2.2 MHz and 4.2 MHz. The operating frequency is set to 3.25 MHz, which yields a penetration depth around 20 cm and a recovery time of the echo (at 120 dB) of 0.25 ms.

A suitably delayed impulse excitation of these 64 transmitter-receiver elements generates a deflected ultrasound beam. The angle between the beam direction and the rotation axis is noted θ. The whole angular sector scanned by one hundred of these consecutive beams is about 60° ($-30° \leq \theta \leq 30°$).

Moreover, while scanning a sector, the probe turns. Thus, the images obtained are not perfect plane sections, but conic sections of the LV (see Fig. 3). The curvature of the surface depends on the ratio, r_c, between the rotation and the scanning speed (respectively $\frac{\Delta\phi}{\Delta t}$ and $\frac{\Delta\theta}{\Delta t}$) as shown

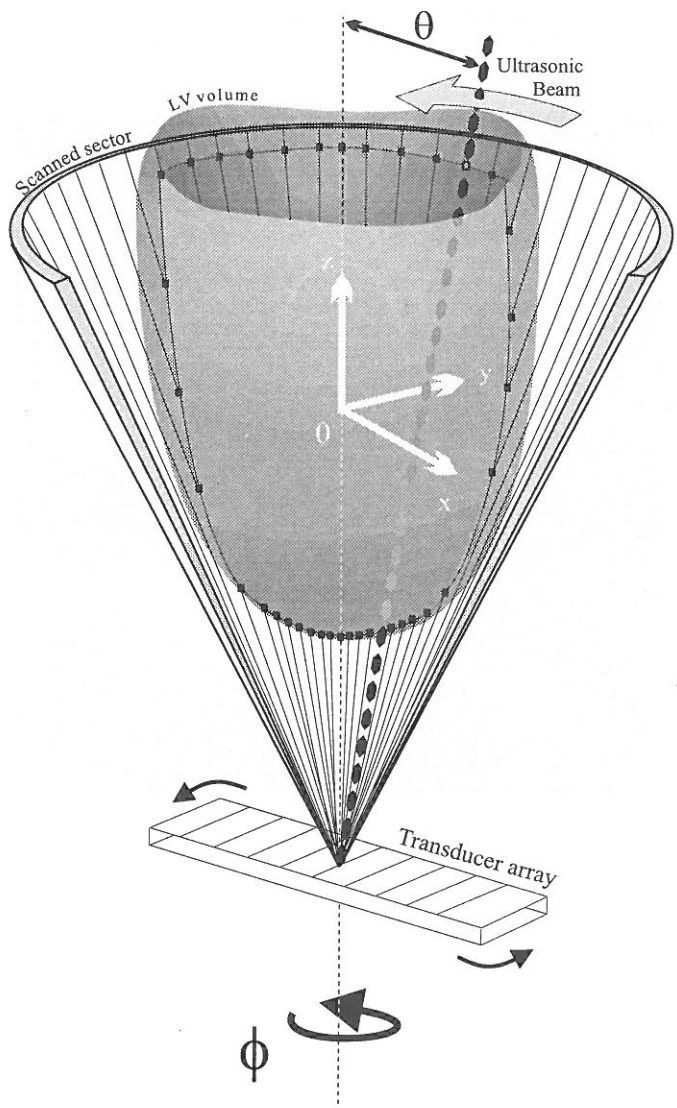

Fig. 3. Schematic view of the sector scanned during one acquisition. ϕ is the rotation angle, θ is the beam angle.

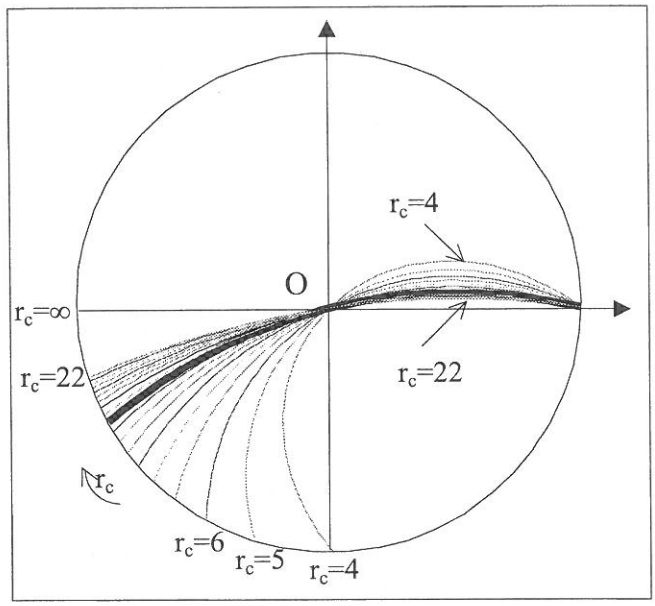

Fig. 4. Curvature of the scanned sector as a function of $r_c = \Delta\theta/\Delta\phi$, where $\Delta\theta/\Delta t$ is the scanning speed, $\Delta\phi/\Delta t$ is the rotation speed. The bold line represents a typical curvature of the sector.

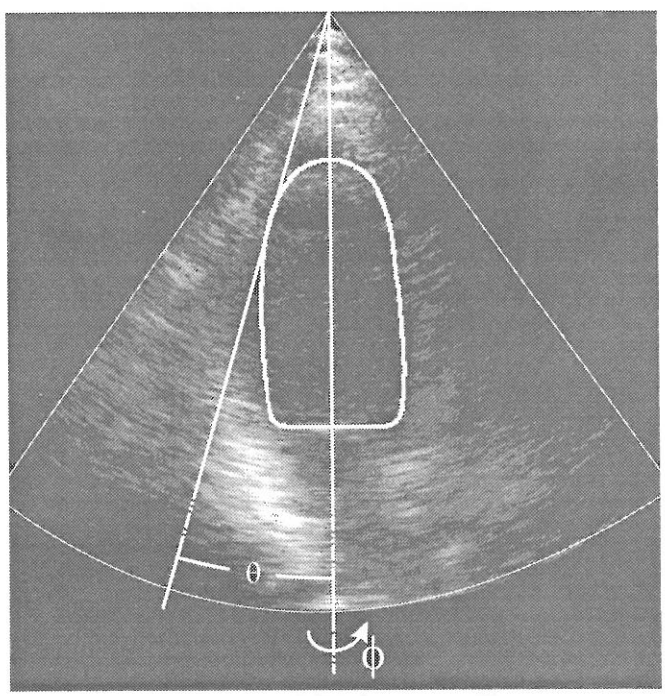

Fig. 5. Example of an ultrasound image recorded during the probe rotation. θ is the beam angle. ϕ is the rotation angle.

in Fig. 4.

1.3. Ultrasound System

The probe is driven by an ultrasound system (ESAOTE AU4) with a specific cardiac module. The system is able to store over 100 consecutive digital images, 508 × 508 pixels, 8 bit per pixel depth. The acquisition frequency is 38 complete 2D-images per second. Figure 5 presents an LV section image with its contour. The practical method used to trace these contours is addressed in the following section.

2. Supervised Contour Tracing

In order to improve the quality of the reconstructed LV volume, and to reduce inter- and intra-operator subjectivity, tracing rules have been defined. The final contour shape and location remain dependent on the physician's interpretation, according to his own medical criteria.

In the present development phase, contours are traced manually according to these rules. In the next phase, only several systolic and diastolic contours will be drawn. These contours will be used as initial conditions for an automatic extraction method based on the harmonic active model (HAM) [Bonciu (1997)].

A complete study of the system *rotating probe – explored volume* has led to the formulation of four tracing rules. They integrate both the physical aspects of our LV ultrasound imaging and the knowledge of an expert : the physician. These four constraints are: *i*) invariance of the local segmentation criteria, *ii*) contour regularity, *iii*) spatial-temporal deformation regularity, *iv*) spatial–temporal coherence.

2.1. Invariance of the Local Segmentation Criteria

The continuous and fast acquisition means that patterns localized in successive images keep their relative locations in each image. In particular, the local criteria used to trace LV contours must be maintained over successive images. On the other hand, the displacements of some reference patterns of images lead to a displacement of the contour relative to the image.

For example, in Fig. 5, the bottom of the mitral valves appears rather bright. This pattern can be used to fix the bottom of the contour of the LV. If the first traced contour is above the bottom of the mitral valves, the relative continuity of images imposes to conserve this criterion for all the images in the sequence.

2.2. Contour Regularity

In apical incidence, which is the case here, the internal volume of the LV can be compared to a semi–ellipsoid. This approximation is usually adopted in ultrasound imaging to model the LV [Seema et al. (1995)]. Contours obtained by cutting a semi–ellipsoid by conic surfaces, correspond to whole or truncated ellipses. Therefore, without being strictly elliptic, the shape of hand-traced contours must be smooth.

This constraint is checked by modelling each contour by a minimal number K of complex parameters. It has been shown [Léger et al. (1994)], [Bonciu et al. (1996)] that the model parameters correspond to the first Fourier coefficients of the polar contour expansion, around an origin from which every radius crosses the contour in only one point. This origin, called central origin, is defined as the center of the largest circle that approximates the contour; in the least squares sense. This number of parameters directly defines the smoothness of the contour, and their values give its shape.

The obtained contour can be very easily deformed by adjusting the values of the parameters interactively. Each deformation is done in an angular sector inside which all the samples are automatically moved in order to minimize the oscillations of the new contour [Bonciu et al. (1996)]. In the present case, $K = 32$ complex Fourier coefficients provide $M = 64$ ($M = 2K$) characteristic samples which, by circular Shannon interpolation, give a smooth and continuous contour on each image. The same technique is also applied in the HAM modelling, where an active polygonal model drives the deformations in each local sector.

2.3. Spatial-Temporal Deformation Regularity

This rule quantifies the amplitude of deformations between two consecutive contours. Given that the LV is continuously deformed with respect to its center of inertia, which moves continuously during the cardiac cycle, the amplitude of deformations must be limited. These deformations have both temporal and spatial origins.

Firstly, regularity is a function of acquisition frequency in relation to the LV motions. Given that two consecutive contours correspond to two different acquisition times, a too low acquisition frequency may increase the amplitude of deformations.

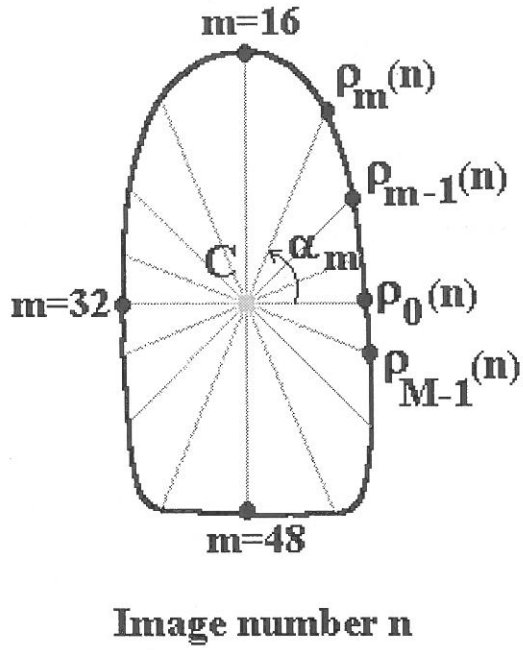

Fig. 6. Position of M characteristic samples along the contour number n and their respective radii $\rho_m(n)$, $0 \leq m \leq M = 64, 0 \leq n \leq N$.

Moreover, the probe rotation is an additional parameter which can significantly modify the regularity. Indeed, consecutive contours which are not acquired in the same angular positions are not assumed to be similar. The amplitude of the recovered deformation was quantified using an ellipsoidal LV model, as a function of the probe movement parameters (speed rotation, position of the rotation axis...) [Bonciu (1997)].

The method used to verify temporal regularity is based on the time revolution of the M characteristic contour samples. Contours traced from the N images are developed around their central origin (x_c, y_c) and the $N \times M$ radii, $\rho_m(n)$, corresponding to the direction α_m, are analyzed. The index n indicates the position of the contour in the sequence, and the index m refers to the characteristic sample along each contour (see Fig. 6).

In each direction α_m, the time evolution of the $N-1$ increment, $\rho'_m(n) = \rho_m(n+1) - \rho_m(n)$, is computed. Then the mean μ_m and the variance σ_m of ρ'_m are estimated. The σ_m value is an indicator of the temporal regularity, in the direction α_m. The contour tracing error may be neglected when the computed value, σ_m, is less than a few inter–pixel distances, 2 or 3 typically. On the other hand, a value $|\rho'_m(n) - \mu_m|$ much greater than σ_m (e.g. $4\sigma_m$) indicates a temporal regularity breakdown of the contour n, in the direction α_m.

Once the outliers are removed, the temporal continuity can be further improved by modifying the contours interactively in order to reduce the value of σ_m in each direction α_m. By reducing the distance between the samples $\rho_m(n)$ on neighbor radii, the overall variance, σ_m, is reduced. Thus, the temporal regularity between the contours is improved. In practice, it is difficult to check the temporal regularity in all the directions α_m at the same time. That is why the horizontal, (α_0, α_{32}), and vertical, $(\alpha_{16}, \alpha_{48})$, directions are inspected first (see examples in section 4).

2.4. Spatial–Temporal Coherence

This rule assumes that the set of images, and consequently the deformations of the LV, are periodic. This assumption is based on two hypotheses. First, the cardiac cycles are assumed to be periodic during the four cardiac cycles needed for reconstruction. This hypothesis is

verified *a posteriori* by using the ECG signal. This is the usual hypothesis in the clinical investigation case and is less restrictive for fast acquisitions. Our preliminary studies on single cycle fast acquisition are encouraging, opening up the future possibility of studying even clinical cases with pronounced arrhythmia.

Second, the probe is assumed to be motionless relative to the LV. The short duration of the examination ($<$ few seconds), and the patient maintained in apnea are sufficient to guarantee this hypothesis. The hypothesis allows the spatial superposition of successive cardiac cycles, and thus, a better exploration space wrapping. In the single cycle acquisition method, which is our final goal, this hypothesis will also become unnecessary. Besides, these two hypotheses are enhanced by using a reduced number of cardiac cycles for the LV reconstruction.

The spatial–temporal coherence can be verified based on these hypotheses. The measures of the rotation angle ϕ are obtained within a normalized temporal cycle ($0 \leq \tau \leq 1$) on the angular interval ($0° \leq \phi \leq 180°$). In this representation, ϕ is limited to 180° because only half a rotation is needed to scan all the volume of the ventricle. Mapping all the acquired data on a single normalized cycle, images exhibiting similar spatial–temporal co-ordinates (i.e. similar ϕ and τ) are better outlined. As a result of the previous periodicity hypothesis, their contours must have similar deformations to be considered as valid data (see example in section 4).

2.5. Supervised Tracing Method

In addition to the previous four tracing rules, it is essential that the physician has the possibility to trace the contours following his own expert criteria. In apical incidence, many parts of contours do not appear because they are lost in noise (next to the apex, for example), not visible (when the walls are not reflexive and parallel with the echo beams), or do not even exist (when the mitral valves are wide open). Only an expert can therefore trace the relevant contours. Moreover, by allowing the specialist to trace the contour he wants, he can select endocardial or epicardial contour of the LV, including the papillary muscles or not. As it is unreasonable to expect the same results from different experts and even from the same expert at different times, all the previous hypotheses have been included in a tracing protocol.

First, the specialist carefully traces the contour of the LV of his choice in the images. Then, the four constraints are checked. When a contour does not meet one of these rules, the expert adjusts it. For this purpose, a sample is moved and this change is extended to the neighbouring samples, spatially as well as temporally. This is done using spatial-temporal weighting windows, either linear or Gaussian. After several trials, the expert becomes familiar with the spatial-temporal constraints, and fewer and fewer corrections are needed.

With the proposed method, the obtained 2D contours are spatially and temporally smooth and periodic with respect to the normalized cardiac cycle. Used to improve hand traced contours, the method may be combined with an automatic detection technique, which should provide similar results [Chalana et al. (1996)], [Cohen et al. (1993)], [Bonciu (1997)]. The obtained contours are used in subsequent processing stages, when, applaying a harmonic modelling technique, the movement and deformation of the LV are obtained.

3. 4D Restitution of the Left Ventricle

The 3D reconstruction of the LV uses the cardiac cycle periodicity hypothesis as explained in the previous section. Thus, the whole set of data is mapped onto one normalized cardiac cycle period.

To reconstruct the LV volume, it is assumed that this remains stationary during a very short period of the normalized cardiac cycle. This allows several images, acquired in the same normalized time interval $\Delta\tau$ in successive cardiac cycles, to be considered as spatially coherent. Moreover, if the deformation and movement of the LV is neglected for this time interval, the image set may be used to obtain an instantaneous 3D model. All the contours belonging to the same temporal window $\Delta\tau$ are then used to reconstruct one volume.

Considering for each instant τ the temporal window $\Delta\tau$ centered on τ, we can obtain the LV model for all normalized cardiac cycle instants.

The width of the window gives the number of images used to render each volume. The wider the window, the greater the number of contours used to render the volume. Thus, the spatial accuracy increases. However, this implies the use of large temporal windows, and the stationarity hypothesis for the LV becomes unrealistic. Therefore, the window width must be chosen in such a way that deformations of the LV are small, while providing a maximum number of contours.

In the present study, a temporal window whose width equals 5% of the total duration of the cardiac cycle is chosen. This is equivalent to assuming that the LV remains stationary during the instants equal to one twentieth of the cardiac cycle.

The samples obtained in the temporal window $\Delta\tau$ are transformed in the absolute co-ordinate system, applaying the probe real rotation law and the scanning beam movement law. This set of data points is used to estimate the local co-ordinate system of the LV (object-centered co-ordinate system). The procedure follows the Chen approach [Chen et al. (1994)], which proposes the ellipsoid as a global LV model. The ellipsoid geometric parameters define the local spherical co-ordinate system (ρ,θ,ϕ).

The samples, $\rho(\theta, \phi)$ which stem from the local model, are irregularly spaced in this co-ordinate system, due to the sequential characteristic of the acquisition device. An interpolation technique, based on the 2-D discrete Fourier transform, is applied to complete the local co-ordinate system grid. The algorithm is based on an *iterative* spectral filtering technique H3DM, which progressively forces the harmonic model to map the original samples [Bonciu (1997)]. This modelling technique makes a mild assumption about the LV characteristics; namely that the LV must have a known band-limited spatial harmonic representation. Once the harmonic local model is obtained, the estimated data points are superposed to the global model (the ellipsoid) and the whole LV is visualized.

The same model is applied for successive temporal windows and the whole LV movement and deformation are obtained. The global movement is given by the evolution of the center of the ellipsoidal model. Global deformation are obtained from the time evolution of the ellipsoid geometrical characteristics. The local movement and deformation are obtained from the H3DM model parameters at each discrete instant. The above approach is well suited for modelling objects with spherical symmetry, i.e. objects which admit a function as representation model in the global spherical co-ordinate system.

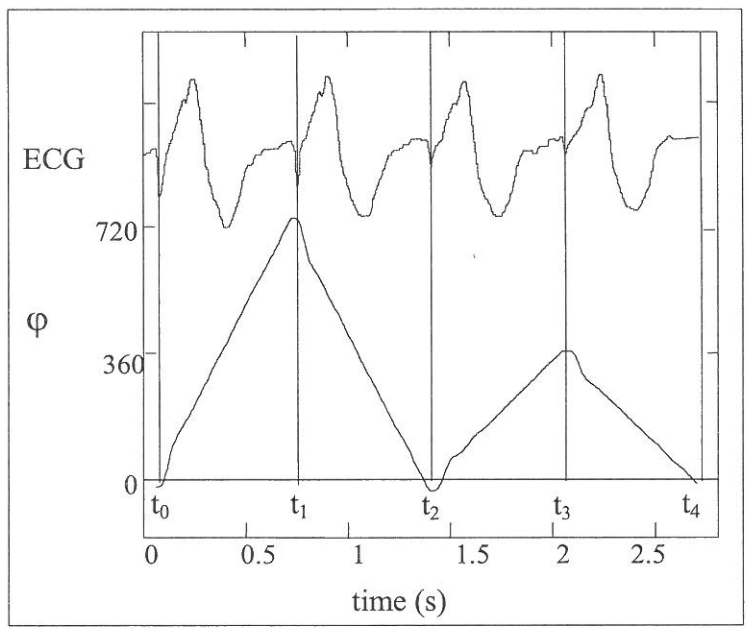

Fig. 7. ECG signal mapped with the rotation angle ϕ, of the probe as a function of time. $t_0, \ldots, t_4$ note the successive cardiac cycles.

4. Experimental Results

The clinical experiments were carried out at the cardiology department of the Regional Hospital in Orléans (France). Recordings were made during 4 consecutive cardiac cycles, for a heart driven by a pacemaker device(T_c = 662 ms) (see Fig. 7). Other specific experimental conditions are detailed in the following section and an LV reconstruction example is presented in the last section.

4.1. Experimental Conditions

As shown in Fig. 7, the probe rotation speed was synchronized with the cardiac cycle. The vertical lines in Fig. 7 indicate the instants $t_0, \cdots, t_4$ of the beginning of the QRS complex rising front. These instants were measured with a 1 ms accuracy. t_0 was chosen as time origin, and also defined the origin (0°) of the azimuthal angles ϕ. During the first two cycles, the sensor rotated 720° per cycle, and the rotation was inverted at instant t_1. During the next two cycles, the rotation speed was halved. From instant t_2, the probe hence rotated 360° per cycle, and the rotation was inverted at instant t_3. The rotation speed changes can be used to improve the angular data distribution in the (θ, ϕ) modelling space. This ensures a greater accuracy

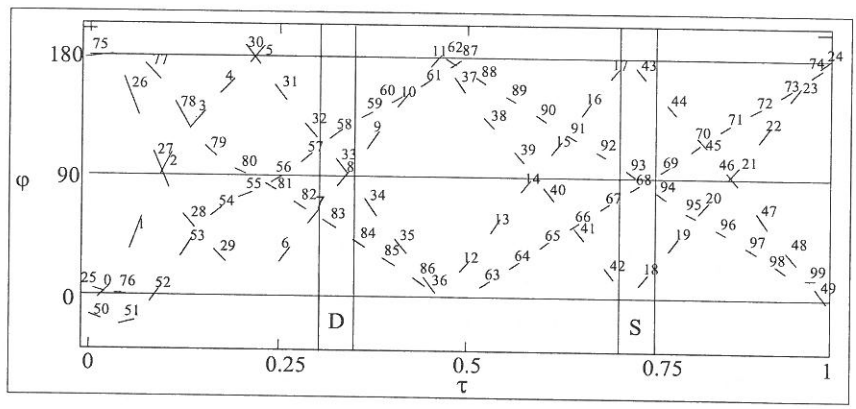

Fig. 8. Angle of rotation of the probe ϕ ($0° \leq \phi \leq 180°$) in normalized time τ ($0 \leq \tau \leq 1$). The set of scanned sections is plotted over the normalized cardiac cycle. Each dash marks the used sector, $\Delta\phi_n$, scanned during the acquisition of the image number n; the point indicates the instant τ_n when the ultrasound beam is aligned with the rotation axis of the probe ($\theta = 0°$). The window D (respectively S) represents the end-diastole (resp. end-systole).

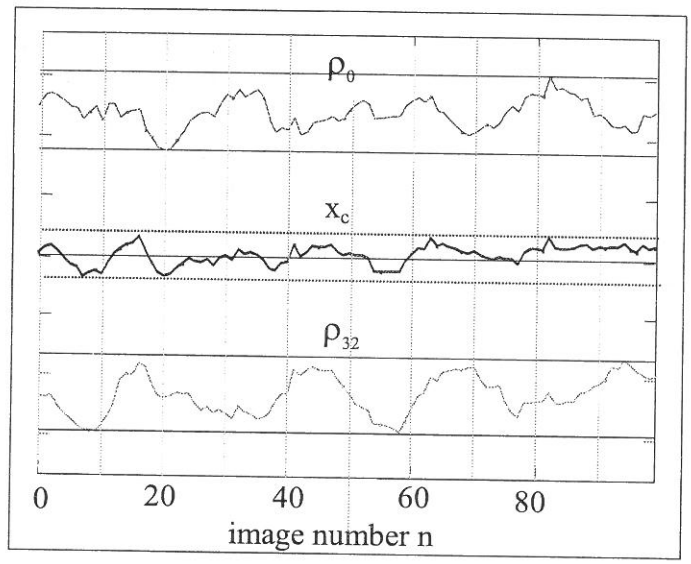

Fig. 9. Time variations of horizontal radii $\rho_0(n)$ and $\rho_{32}(n)$, as well as the abscissa $x_c(n)$ of the central origin.

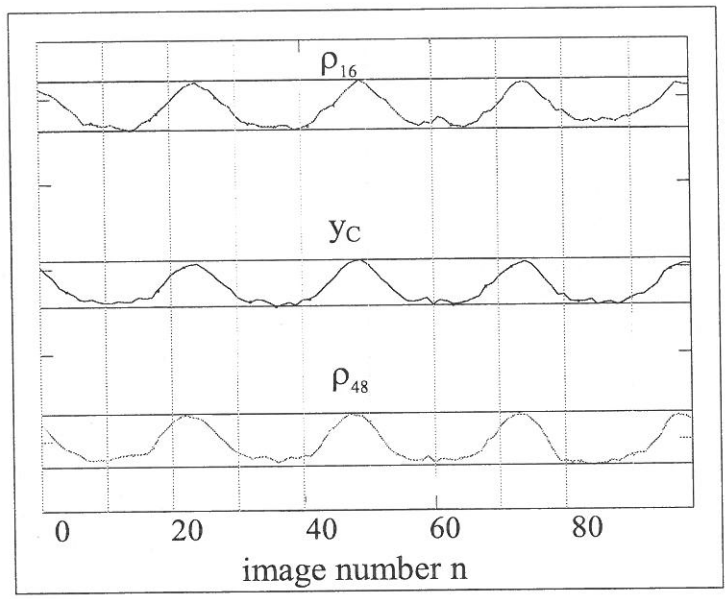

Fig. 10. Time variations of vertical radii $\rho_{16}(n)$ and $\rho_{48}(n)$, and the ordinate $y_c(n)$ of the central origin.

of the H3DM modelling technique and also reduces the computational complexity (the number of iterations for uniform data distribution modelling with H3DM is minimal). As it is shown in Fig. 8, the speed variation presented in this paper is not optimal. Indeed, certain cardiac instants ($\tau = 0$, $\tau = 0.5$) are not very well defined spatially. This can be fixed either by increasing the temporal window $\Delta\tau$ or by adopting another speed configuration.

Figure 9 shows the time variations of the horizontal radii, $\rho_0(n)$, $\rho_{32}(n)$, as well as the abscissa of the central origin, $x_c(n)$, of the consecutive contours. It is very sensitive to the rotation of the probe, whereas the transverse diameter $|\rho_0(n) - \rho_{32}(n)|$ of the LV is directly obtained at each instant of acquisition. The regular LV motion can be observed. Similarly, Fig. 10 shows the time variations of the vertical radii, $\rho_{16}(n)$, $\rho_{48}(n)$, as well as the ordinate of the central origin $y_c(n)$. The ordinate evolution is quite insensitive to the rotation of the probe, since, during acquisition its axis is aligned with the LV axis (maximal deviation is less than 10°). Moreover, the LV motion is not observable, because the vertical diameter $|\rho_{16}(n) - \rho_{48}(n)|$ does not change significantly. Nevertheless, the periodic motion from high to low of the LV during the 4 cycles can be observed.

4.2. Experimental Results

In the above experimental context, the spatial-temporal reconstruction method leads to the LV model in Fig. 11. The 2D contours used in this reconstruction are reference contours traced by the expert, using the spatial-temporal coherence constraints.

Further analysis of the 2D contours and 3D geometrical model parameter variations was performed. First, the regularity of the deformations was checked *a posteriori* by comparing images at the same angular location for the same normalized instants in different cardiac cycles. Second, other independent cardiac measures were obtained for the same patient. One of these clinical parameters is the ejection fraction, obtained from the end-systolic and the end-diastolic volume measurements. The end–diastolic volume V_d is an important parameter, which indicates a volumetric overload or an abnormal myocardial contractility. The end-systolic volume V_s is a derived parameter, which depends on the end-diastolic volume, on the myocardial contractility and on the volume overload. The systolic ejection volume and the ejection fraction can be estimated from the measurements of the V_d and V_s, (see Fig. 12) following the equations :

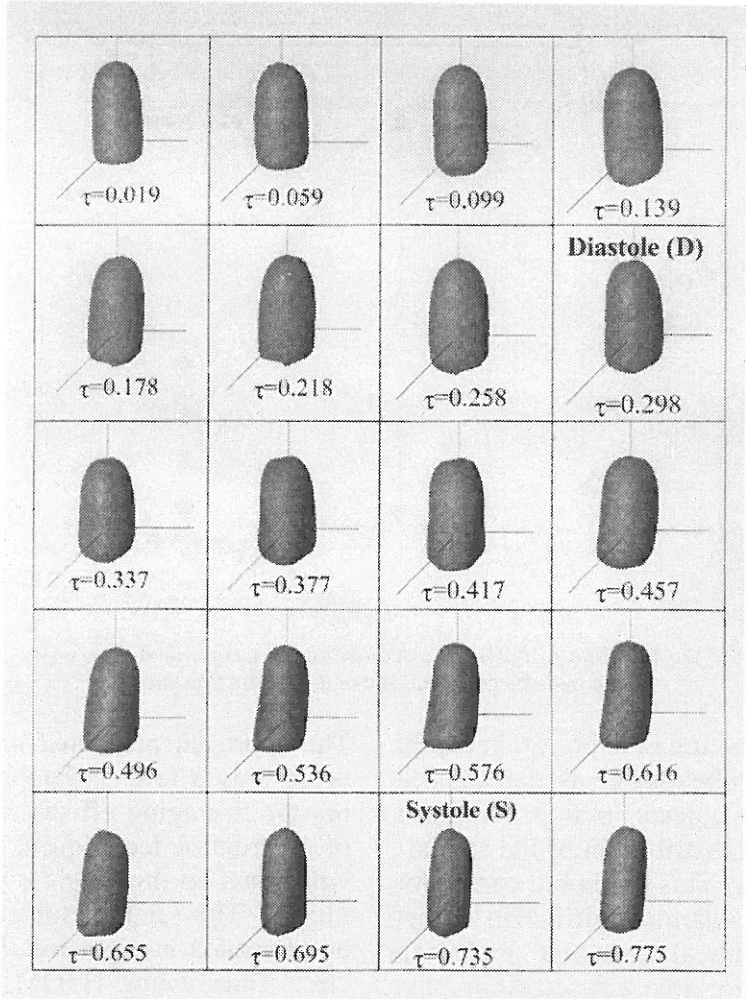

Fig. 11. Examples of the rendered LV volumes. τ represents the normalized time ($0 \leq \tau \leq 1$) in relation to the cardiac cycle duration.

$$V_{se} = V_d - V_s,$$
$$F_{ej} = \frac{V_{se}}{V_d}.$$

Independent measures of F_{ej} show results in good agreement with the measure obtained from the 3D reconstruction. ($F_{ej} = 0.416$). This value corresponds to a hypokinesia of the ventricular wall, caused by a coronary thrombosis, as the other clinical investigations have shown.

This experiment confirms the efficiency of the proposed reconstruction method. The reference 2D contours were used to estimate the cut-off frequency of the H3DM model, and to test its capabilities on real clinical data.

5. Conclusion

This paper presents some preliminary results obtained in the LV spatial-temporal reconstruction, using ultrasound images acquired during only four consecutive cardiac cycles. A new trans–thoracic probe has been developed for this purpose. The probe rotates continuously and acquires apical images of the LV, at different angles. In the present study, the rotation has been synchronized with the cardiac cycle, in order to acquire spatially and temporally periodic images.

The probe is specially designed for high-frequency acquisition systems, and does not require recovery time as is the case with step-by-step movement probes. The acquisition frequency is limited only by the physical propagation time of the ultrasound beam through the

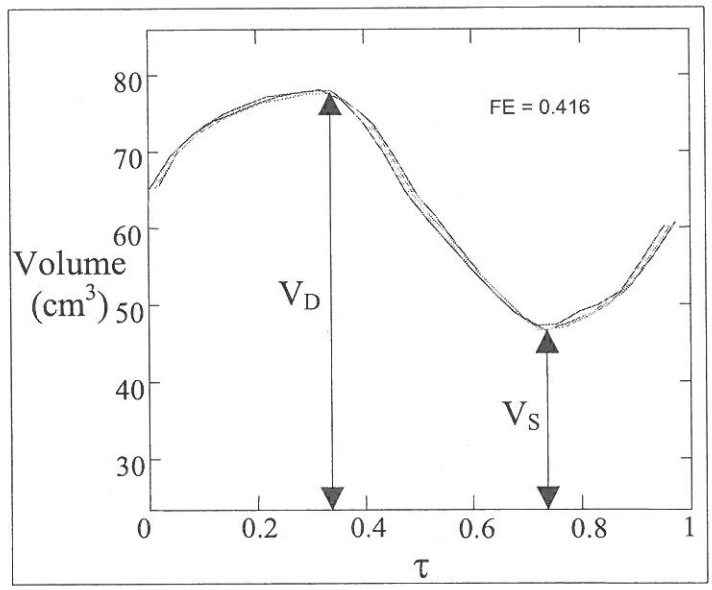

Fig. 12. Time evolution of the LV volume, during the acquisition, in normalized time τ ($0 \leq \tau \leq 1$). V_D and V_S are the end-diastolic and the end-systolic volumes.

tissue, which is a function of exploration depth. Consequently, the probe may be used to acquire data on deformable objects, thus providing a limit on the spectral distribution of the spatial-temporal data points. This harmonic constraint is, however, a mild constraint, fulfilled in almost all the domains of clinical ultrasound, including the targeted application area, echocardiography.

A specific harmonic modelling technique was developed to cope with the geometrical complexity of the reconstruction problem, which produces a non-uniform data distribution in the reconstruction space. In this paper, an H3DM model is briefly discussed and applied to a reference data set.

In order to extract the expert information and to use this type of information effectively in the modelling process, several coherence criteria have been derived. These criteria have been transformed into coherence constraints in the clinical investigation protocol. These constraints are imposed on the experts, in order to minimize the inter- and intra-observation variability during the contour tracing procedure, a problem often encountered in medical image analysis. Moreover, the spatial-temporal constraints may be considered here as an additional tool in helping the expert detect the abnormalities, when the constraints are fixed to values corresponding to a normal heart.

The approach presented has the advantage of using only a few cardiac cycles and of avoiding the averaging effects encountered in other reconstruction techniques. Our future research will focus on the single cycle modelling techniques. The rotation speed of the sensor must be increased to eight rotations per cardiac cycle. Thus, using H3DM three LV contours, angularly separated by 60° would be sufficient to obtain the instantaneous LV model.

When the processing time is significantly reduced, the modelling technique will be supervised by the expert. The expert will be able to adjust the resulting volume directly, when the image quality is not sufficient for an accurate contour detection, or when the detection does not agree with the medical criteria. Finally, when the above improvements are effective, the study will be carried out on patients with significant arrythmia, using a single cardiac cycle.

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