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Model of probe motion in transthoracic 3D-echocardiography for assessing the precision in left ventricular cavity volume measurement

M. J. Ledesma-Carbayo[a], J. Declerck[b], J.A. Noble[b] , A. Santos[a]

[a] ETSI Telecomunicación, Universidad Politécnica de Madrid, 28040, Madrid
[b] Department of Engineering Science, University of Oxford, Parks Road, Oxford OX1 3PJ, UK.,

Abstract. Over the past few years, 3D cardiac ultrasound techniques have been actively developed in research laboratories. Providing a good trade-off between image quality, spatial sampling and rapidity of acquisition remains a major issue to be solved in order to bring this powerful tool into routine clinical practice. A rotating 3D probe allows a reasonable quality 3D image of the heart in motion to be acquired rapidly, with a good spatial sampling. However, because the acquisition is performed without a position sensor, some inevitable motion of the probe occur during the acquisition. The aim of this article is to model and analyse the influence of the motion of the probe on a quantitative measurement (left ventricular cavity volume). A positioning information tracking system is attached to the probe to recall information about the movement induced by the user in a normal acquisition. From this data the noise is modelled and simulated to compute the effect on the volume measurements derived from a software phantom. The effect of probe motion is proven to be below 2 percent of the LV volume under normal conditions of acquisition.

1 Introduction

Ultrasound is the most broadly used modality in clinical practice to assess the function of the heart. Current clinical protocols all require a cardiologist to evaluate visually the quality of the motion on conventional 2D slices of the heart: short-axis, two, three and four chamber views. Although these protocols are well established in modern medicine, the spatial coverage of the assessed areas is low (three views) and the evaluation is purely qualitative. Recently, three-dimensional ultrasound methodology has been introduced to provide a way to acquire image data over the whole heart. In the literature, the different techniques have been classified into three technology generations [1]: first, the purely free-hand system, whereby a position sensor is attached to a conventional 2D probe, and as many 2D slices as necessary are acquired in order to cover as much as possible of the area of interest. The second generation systems use a mechanically driven transducer to acquire a series of 2D planes. The spatial coverage of these probes is more regular, but the motion of the probe can induce erroneous reconstruction [2]. The third generation systems is the only purely real-time three-dimensional modality that exists at the present time: a 2D phased array transducer acquires a volume of data at an image sample frequency in the range from 20Hz to 40 Hz. The spatial coverage is as good as the echographic window allows, but the image quality is still low.

In this article, we used a trans-thoracic probe (second generation type) which samples space with a series of co-axial planes, in our case, aligned with the long axis of the heart. The main drawback of this method is the required stability on the probe position and orientation, and the need to maintain the relative position between the probe and the organ under study. Small deviations on the probe position and orientation may result in misregistration and distortions on the derived surface or the 3D image reconstruction. In the particular case of 3D+T heart acquisitions further problems are involved: the data has to be synchronised with the ECG and a whole cycle is acquired at each transducer plane position, discarding cycles too long or too short. On the whole, the acquisition time is increased significantly (one minute for a typical 12 planes acquisition). This increases the chances of unexpected motion of the probe during the acquisition, as it is not constrained [3]. Patient respiration may also affect probe movement and relative displacement between the heart and the probe, although they could be avoided through respiratory gating but with an additional increase in the acquisition time. All these problems determine the quality of the 3D acquisition and its clinical usefulness.

E-mail: mledesma@die.upm.es, {jdecler,noble}@robots.ox.ac.uk

This paper presents a study carried out to understand the effects of the probe movement on the 3D transthoracic acquisition, the resulting 3D reconstruction and derived quantitative parameters. Using a positioning information tracking system attached to the probe, information about the movement induced by the user in a normal acquisition is obtained. From this data the motion is modelled and simulated to compute its effect on the volume measurements derived from a software phantom.

2 Methods

2.1 Real data acquisition

Transthoracic probe and positioning information system

Digital 3D echocardiographic data was acquired on a HP Sonos 5500 ultrasound machine using a 3-5 MHz 3D transducer (Agilent Technology, Andover, MA, USA). This probe is composed of a phased array transducer that is rotated by a stepping motor around the main axis of the probe from –90 to +90 degrees. For the experiment, the probe is attached to a 6 degrees of freedom multi-axis articulated arm (FaroArm, Faro Technologies, CA, USA) to register the motion (translation, rotation) of the hand of the sonographer holding the probe during the whole acquisition. The mechanical arm features were chosen carefully in order not to restrict the operator movements, to have a high registering rate and good accuracy. The mechanical arm is connected to a computer that controls the position information capture. A conventional 11 positions calibration process is needed before starting the positioning capture to relate the tip of the probe with the arm reference system. Position information and image acquisition are synchronised manually, as it is impossible to get from the probe or the machine the exact time at which each image is acquired.

Acquisition protocol and data reconstruction

Different operators performed several exams on voluntary patients. During each examination, around 20 frames were acquired at each plane position and stored on a disk as 2D echograms (12 coaxial planes separated by 15 degrees and 21 frames at each position per cardiac cycle). The number of frames depended on the heart rate of the patient. The images were acquired at a fixed frequency of 25 Hz. All acquisitions were performed from the apical view direction (the 3D probe aligned as much as possible with the LV long-axis).

Once the position information and the image data were correlated, the image planes were automatically repositioned in space to evaluate the deviations from the ideal acquisition pattern.

2.2 Probe motion simulation

Probe motion model

The motion of the probe is the only one considered here. That is we do not take into consideration the motion of the organ due to respiration, for instance. The model follows some observation on the actual protocol which is used by a sonographer to acquire the image: a sonographer holds the probe in their best hand and tries to stay as still as possible during the one or two minute acquisition time. Let us call S the direction of the forearm, T the direction of the probe (the long axis) and N a vector such as T x S = N (Fig. 1). The probe is held from the tip of the fingers for better control, the direction of the probe (therefore, the common axis of all image planes) is actually perpendicular to the arm direction S. The forearm rests on the patient's body for better comfort.

The motion of the probe is modelled as a process of elementary displacements (rotation, translations) happening at regular intervals. Starting the acquisition with an identity (the motion is referred to that starting point), a new elementary displacement D_t is generated and right multiplied to the displacement matrix M to update the new position. The final probe position after a certain time *t* is described by the displacement matrix $M = D_t \, . \, D_{t-1} \, . \, \ldots \, . \, D_0$. In our experimental setup, the position sensor gives a position reading every 30ms (33 Hz): the statistical process will follow the same characteristic.

In more detail, each displacement D_t is decomposed as follows: the motion around the S axis is quite easy, as the forearm can roll on the patient's body. The motion around the N axis is more difficult, as the wrist has to firmly tighten the probe to the patient's chest. The rotation around the T axis is even more constrained, as it would represent a motion of the probe rolling in the sonographer's fingers, highly unlikely to happen. Finally, some translation can occur as the contact between the probe and the chest is lubricated by ultrasound gel. The skin can also slide on the ribs, adding an extra component to the translation.

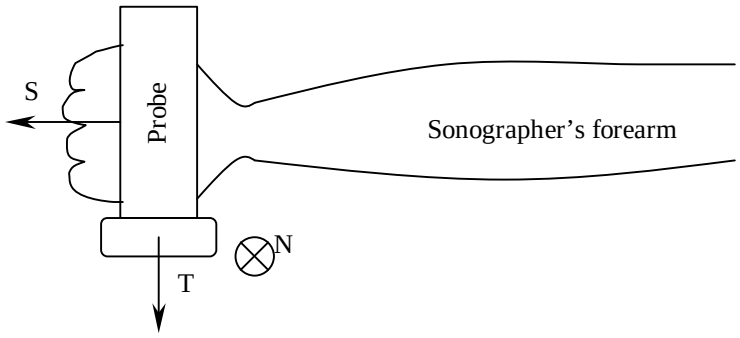

Figure 1. The sonographer as he holds the probe, defining natural axes for the representation of the scene.

Each individual rotation has therefore six components which have clearly identified origins. As a first approximation, we consider these components independent random variables, three for the rotation, three for the translation. Following a classical framework for rotation noise models [4], the random rotation is described as a rotation vector which has three components that follow a Gaussian distribution. The means and standard deviations of these random variables can be estimated from real data.

Volume measurements and reconstruction effects

The acquisition process was simulated over an artificial set of data generated using the probe motion model. The data used was a simple ellipsoid (which has a shape and position similar to that of a left ventricle). The volume was computed using a classical method to calculate volume from a sample of radial slices: each slice is intersected with the volume of interest, generating an ovoid shape. The contribution of this slice to the volume is calculated by integrating voxel mass weighted with the distance of this voxel to the common axis of the planes. This elementary volume corresponds, in three-dimensions, to a "slice of a pie". The total volume is the sum of all elementary volumes computed from each slice. The change in volume was measured as the motion parameters were varied.

Five real datasets were acquired in order to validate the noise model and to be able to know, for a typical acquisition, what the error on volume measurement can be. These datasets were acquired only to measure the motion of the probe, not to measure the volume of a real left-ventricle, as we did not have the capacity to validate this measure.

3 Results and discussion

3.1 Extraction of probe motion model characteristics

Parameter	μ	s	KS *Z*	KS *P*
Rot S (rad)	$3.8\ 10^{-6}$	$8.3\ 10^{-4}$	0.419	0.995
Rot N (rad)	$1.8\ 10^{-6}$	$6.0\ 10^{-4}$	0.593	0.873
Rot T (rad)	$-2.9\ 10^{-6}$	$4.5\ 10^{-4}$	0.474	0.978
Tr S (mm)	$-1.3\ 10^{-3}$	0.22	0.441	0.990
Tr N (mm)	$-2.4\ 10^{-3}$	0.30	0.535	0.937
Tr T (mm)	$9.8\ 10^{-3}$	0.25	0.740	0.644

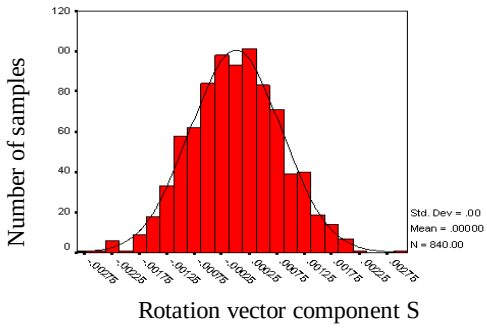

Table 1. Right, means and standard deviations for the parameters as extracted from the real data. P values of the Kolmogorov-Smirnov (KS) test show how valid the Gaussian approximation is. Left, an example of a histogram of values for the rotation angle S and the corresponding Gaussian curve overlaid.

From the position sensor data, we computed the elementary rotations R_t. Typically, each experiment gave around 1000 measurements. The characteristic means and standard deviations of the translational and rotational components of all R_t were extracted. A Kolmogorov-Smirnov test was performed to check how valid was the hypothesis that the random variables are Gaussian with zero means. Table 1 shows the values means and standard deviations obtained for the first acquisition (for instance). *P* values of the KS test show that this assumption is correct.

3.2 Volume measurements using the probe motion model

The parameters of the model (standard deviation of Gaussian random variables) were set to some values by averaging all the values obtained from the five acquisitions. To simplify the assessment of volume measurement error induced by probe motion, the standard deviations were expressed as multiplicative factors of a generic variable σ_M, the other parameters of the model being derived according to Table 2. Values were multiplied uniformly by X = 0.5, 1, 1.5, etc. to set a level of motion of the probe. A value of 2, say, means that the values used for the model were all twice the values obtained from the characterisation. A Monte Carlo simulation using 50 trials per value of X was performed. The result of these simulations are summarised in figure 2: the horizontal curve shows the average error in volume estimation (in mm^3), to be compared to the value of the volume itself (208.10^3 mm^3). The vertical lines represent error bars with limits set at one standard deviation above and below the average. For X=1, this error is less than $3\ 10^3\ mm^3$.

Parameter	Parameter value	Multiplicative factor for σ_M
σ_M	$4.79\ 10^{-4}$	
Rot S (rad)	$7.65\ 10^{-4}$	$1.60^*\ \sigma_M$
Rot N (rad)	$7.50\ 10^{-4}$	$1.57^*\ \sigma_M$
Rot T (rad)	$4.79\ 10^{-4}$	$1.00^*\ \sigma_M$
Tr S (mm)	$2.09\ 10^{-1}$	$437.00^*\ \sigma_M$
Tr N (mm)	$2.73\ 10^{-1}$	$570.00^*\ \sigma_M$
Tr T (mm)	$2.45\ 10^{-1}$	$512.00^*\ \sigma_M$

Table 2. Noise parameter values used in the model of probe motion. Random variables have zero means, their standard deviations is calculated by multiplying σ_M with a constant factor.

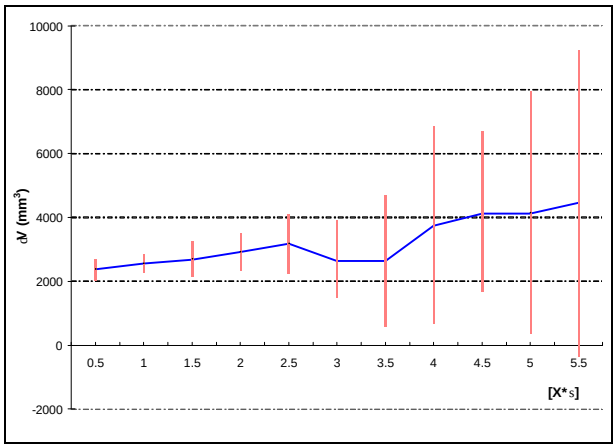

Figure 2. Volume measurement deviations vs-different multiplications of σ_M.

4 Conclusions

The simple motion model of the probe gives an estimation of the intrinsic precision in volume measurement using a trans-thoracic 3D probe. It has been shown that the volume can be estimated within 3 ccm for normal acquisition conditions. This value is below 2 percent of the LV volume.

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