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● *Original Contribution*

VALIDATION OF A VOLUMIC RECONSTRUCTION IN 4-D ECHOCARDIOGRAPHY AND GATED SPECT USING A DYNAMIC CARDIAC PHANTOM

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Abstract—A dynamic cardiac phantom was used as a reference to compare the volumes reconstructed with 4-D echocardiography and gated single-photon emission computed tomography (SPECT). 4-D echocardiography used a new prototype of rotating scan head to acquire ultrasound (US) images during a cardiac cycle, associated with a new protocol (left ventricular 4-D or LV 4-D) to reconstruct the volume deformations of the heart as a function of time. Gated SPECT data were acquired with a standard single-head gamma camera, and the reconstructions were carried out using the Mirage software released by Segami. The influences of different LV 4-D parameters were tested and analyzed. End-diastolic volume, end-systolic volume, and ejection fraction were measured using both LV 4-D and gated SPECT. Results obtained showed a straight correlation between the two examinations. The agreement confirmed the relevance of the comparisons. This study is an initial step before conducting clinical trials to exhaustively compare the two modalities. (E-mail: Long-Dang.Nguyen@chr-orleans.fr) © 2003 World Federation for Ultrasound in Medicine & Biology.

Key Words: **Dynamic cardiac phantom, Quality control, Myocardial gated SPECT, 3-D/4-D echocardiography.**

INTRODUCTION

The human heart is a 3-D organ with a complex shape and a periodic motion. For a correct evaluation of systolic function, a 4-D reconstruction (3-D volumes as a function of time) seems to be necessary. However, no reference method currently exists for cardiac volume reconstruction; doubtless because of the intrinsic complexity and the recent nature of all multidimensional imaging techniques of the heart. Most of the studies carried out compared two or more modalities: for example, echocardiography vs. gated single-photon emission computed tomography, SPECT (Nichols et al. 2000), 3-D echocardiography vs. nuclear magnetic resonance, NMR (Chuang et al. 2000; Bauer et al. 2001) or vs. isotopic ventriculography (Nosir et al. 1998), or nuclear medicine vs. angiography (Yamazaki et al. 1997). Results sometimes pointed out differences of volume quantification, but it is often difficult to select one method in preference to another, without an accepted standard. Calibration of methods using a beating cardiac phantom, with known parameters, is an appropriate response to these difficulties, and a necessity before clinical evaluation.

Echocardiography was, for a long time, limited to a 2-D examination. Real-time 3-D echocardiography (von Ramm 1990) is a promising new technique (Qin et al. 2000, 2002) based on a matrix probe. However, it requires an additional costly system that cannot replace a state-of-the-art ultrasound (US) machine. We developed a new 4-D method called LV 4-D or left ventricle in four dimensions. It uses a fast rotating transthoracic probe connected to a commercial echocardiograph (Canals et al. 1999). Volume reconstruction is carried out using a sequence of images acquired within a single cardiac cycle (Nguyen and Léger 2002). Quantification using a beating cardiac phantom is of great interest to evaluate and quantify the influence of the different parameters used in the reconstruction algorithm.

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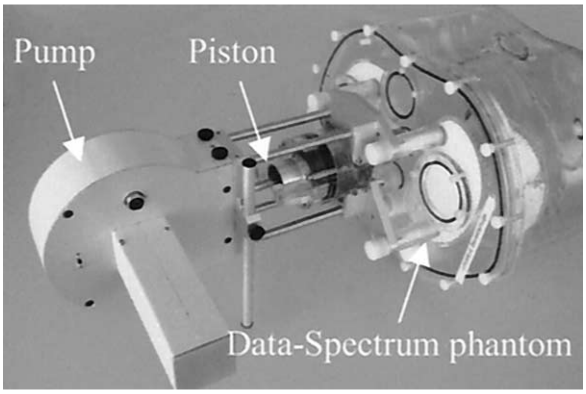

Fig. 1. Dynamic cardiac phantom: full view. Pump motor is in the foreground.

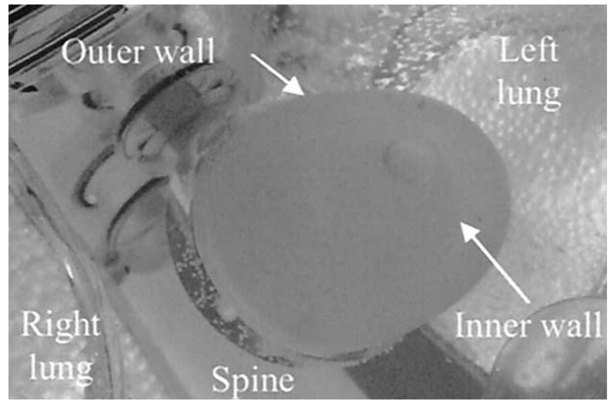

Fig. 2. Details of the two imbricated membranes forming the cardiac cavity.

Gated SPECT is a nuclear medicine imaging method used to evaluate myocardial perfusion using injection of radioisotopes. It is known to be a "gold standard" for volume reconstruction of the heart, and is often used to validate new techniques. Cardiac volumes are computed starting with a set of projection images obtained with a gamma camera; however, results obtained depend slightly on the software parameters used for the reconstruction. A validation using a beating cardiac phantom is essential to optimize the software parameters, and ensure the accuracy of measurements.

The study described in this paper was carried out to compare beating cardiac phantom volumes reconstructed with LV 4-D and gated SPECT. It served to detect and correct errors made within both modalities, by analyzing and adjusting critical software parameters. Results will be used to conduct further clinical comparisons (Debrun et al. 1999) on healthy and ill subjects.

MATERIALS AND METHODS

Phantom description

We used a 3-D gated dynamic cardiac phantom manufactured by the Academic Medical Center of Amsterdam (Visser et al. 2001), shown in Fig. 1 and Fig. 2. It consists of a cardiac insert fitted in the Data Spectrum anthropomorphic torso phantom (Data-Spectrum, Chapel Hill, NC). This insert has flexible silicone walls that form the inner and outer walls of the simulated left ventricle. Water is inserted in the cavity, and the space between the walls is filled with water (echocardiography) or active solution (gated SPECT). At this time, no right ventricle is available to allow two-chamber examinations. The choice of materials (Plexiglas for the torso, silicone for the ventricle, water) simulates clinical imaging very closely. It is suited for nuclear medicine, and acquisitions with US probes are possible, although restricted to a few orientations, including the long-axis acquisition required for LV 4-D. No geometric distortions were noticed in US images of motionless phantom. Shape and size of the simulated ventricle were in good agreement with the phantom.

A pump is used to vary the volume in the left ventricular cavity smoothly, and to produce a realistic stroke volume curve. A fixed 70 mL_3 stroke volume is applied for each ejection fraction, but systolic volume is adjustable, starting from a volume of 50 mL_3. The diastolic volume was varied from 120 to 200 mL_3, and the systolic volume from 50 to 130 mL_3, using steps of 20 mL_3, the corresponding ejection fraction varying from 58% (a value close to the mean normal value) to 34% (pathologic cases). Ten precisely known volumes were obtained, which is sufficient to draw conclusions within the range of available values. Electronics controls the frequency of heart beats (from 60 to 80 beats per s) and delivers a standard electrocardiogram (ECG) signal.

4-D echocardiography

The LV 4-D protocol was developed at the Laboratoire d'Électronique, Signaux, Images (LESI) of the University of Orléans, France. Echocardiographic images, acquired during a single cardiac cycle, are computed to reconstruct the surface deformations of the left ventricle (Bonciu et al. 2001). The innovative feature of the method is the rapid, continuous rotation of the sensor of a classical 2-D phased-array scan head, which, thus, functions as a true 3-D sensor (Fig. 3). This probe is linked up to a commercial cardiac US system HDI 5000 (Philips Ultrasound System, Suresnes, France). Because of the rotation of the sensor, the US beam sweeps a conic surface where the contour of the left ventricle is as visible as on a standard planar image (Fig. 4). The angle α between two successive images is directly related to

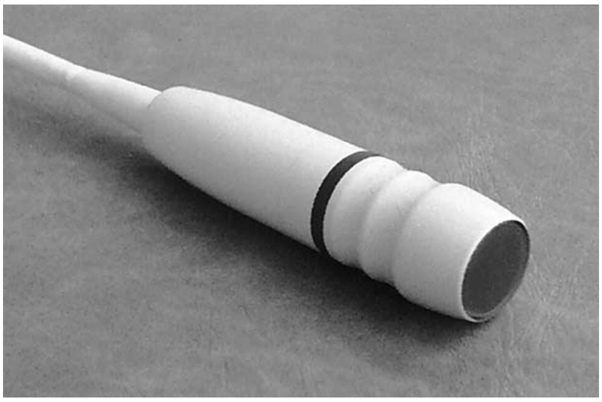

Fig. 3. Prototype of transthoracic rotating probe.

the acquisition frame rate (*FR*, in images per s) (which is not easily adjustable on most US systems) and probe rotation speed (*RS*, in rotations per s):

$$RS = \alpha \times FR/360°. \tag{1}$$

Rotation speed can be discretely adjusted by the piloting software. The contours of the left ventricle are located on each image of the cardiac cycle. The intersections of the endocardial contour and US beam (for example, points I_1 and I_2 in Fig. 4) are described in spherical coordinates by a three-variable function $\rho(\theta,\varphi,\tau)$, relative to an inner

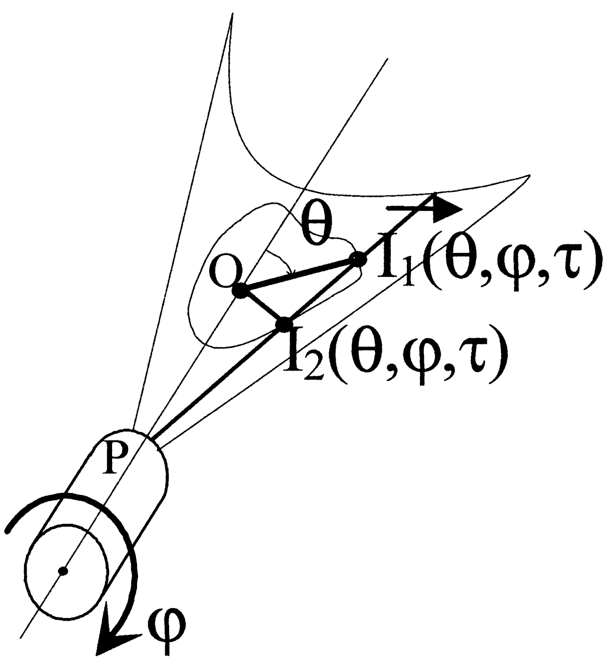

Fig. 4. Example of an image acquired on the cardiac phantom with the rotating probe.

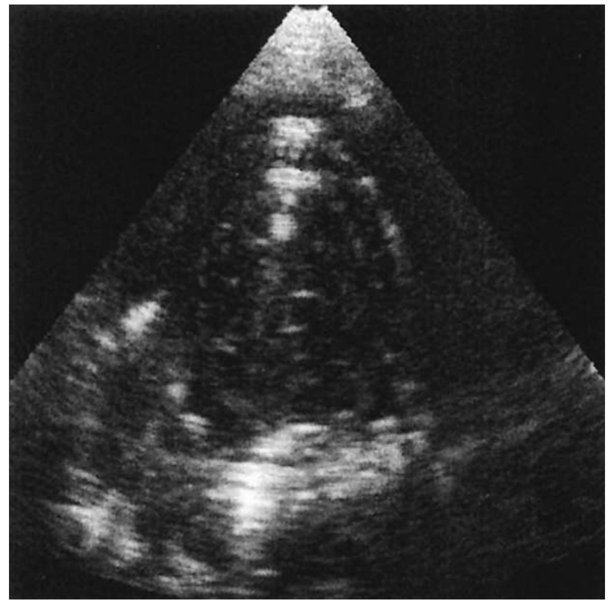

Fig. 5. Principle of operation of the rotating probe to acquire volumic data of the ventricle.

origin. Intersections are defined by an instant of acquisition and coordinates relative to the probe's references: they are real 4-D data.

To reconstruct the LV volumes, the samples of all the contours of the sequence obtained with the rotating probe are inserted in a parallelepiped 3-D grid, the sides of which represent the θ, φ and τ variables (Fig. 6). The 3-D grid, incomplete due to the rotation of the probe during image acquisition, is then interpolated using an iterative algorithm based on the Fourier transform. At each iteration, the 3-D Fourier transform of the grid is computed. Because missing data in the grid contribute mainly to high frequencies in the Fourier domain, a low-pass filtering is applied to reduce discontinuities in the temporal domain. An estimation of missing data is then obtained applying the inverse Fourier transform. To avoid the smoothing of initial data between two iterations, known data from the original grid are retained. The new surface, thus generated, is then used as input for the next iteration and the process is repeated. The algorithm stops when the estimated values do not vary significantly from one iteration to the following, relative to a fixed threshold. This interpolation algorithm is used to distribute the initial information over the entire grid, leading to a uniformly sampled surface that interpolates initial data. The LV surfaces are directly extracted from the interpolated grid, at different instants of the cardiac cycle. Each surface corresponds to a slice $\rho(\theta, \varphi)$ of the grid at instant $\tau = \tau_0$ (Fig. 6).

From a strict signal-processing point of view, the acquisition of numerous samples (several thousands of

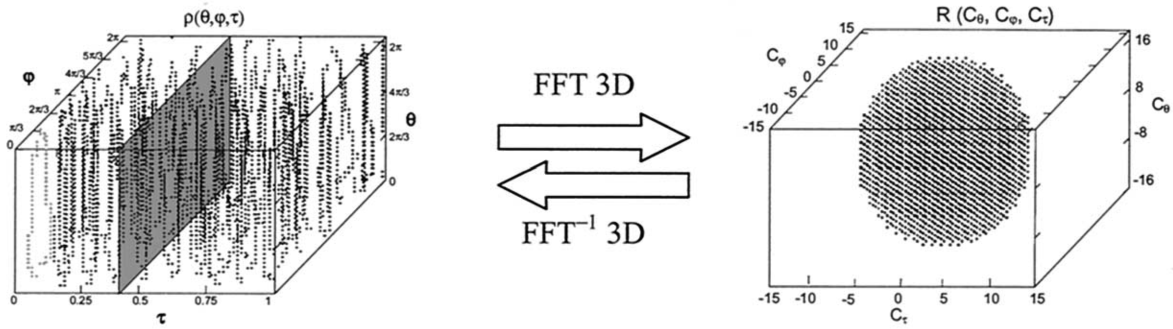

Fig. 6. Interpolation of the 3-D grid with missing data. (left) Dots correspond to the samples of the 3-D LV contours, and the grey plane represents data (original and interpolated) used to reconstruct one LV volume at one instant of the cardiac cycle. (right) Fourier coefficients retained for low-pass filtering.

data, actually) within a signal period (cardiac cycle) allowed us to develop a quadridimensional modeling of the LV volume deformations. By merging spatial and temporal information, this deformable model is well designed for the LV reconstruction, even if the spatial distribution of 3-D data is nonuniform or varies in function of time, which is the case for data acquired with a rotating probe nonsynchronized with the ECG. In practice, the algorithm was included as a plug-in in the echographic image processing software developed by Philips Ultrasound System.

All echographic acquisitions were carried out in fundamental mode. Previous tests using 2nd harmonic mode showed no significant enhancement of image quality on the phantom. The frame rate was set to 53 images per s. About 150 images (320 × 256 pixels) were available for reconstructions (corresponding to 3.4 cardiac cycles for a cardiac frequency of 60 beats per s). Echographic sector was set to 90°, and the depth of examination to 15 cm. Focal point was adjusted at midventricle. For each acquisition, the probe was positioned to ensure the best alignment with the ventricular axis. Acquisitions lasted about 10 s in three stages: a first one with a motionless probe, a second one when the rotating probe accelerates, and a third one with a stable speed of rotation. Typical speed is 8 rotations per s, leading to a 60° angle between two successive images. End-diastolic time is defined by the beginning of the QRS wave, synchronous with the largest volume of the phantom. The first stage is used to determine end-systolic time when the phantom volume is the smallest. The last stage lasted about 4 s, that is to say three full cardiac cycles, all used for volume reconstruction in the study. For each of the cycles acquired, three end-diastolic and three end-systolic contours (approximating volume information over 360° with a speed of 8 rotations per s) were traced by the operator to initiate the reconstruction algorithm. After less than 1 min of computing, 3-D beating volumes are available (Fig. 7). Volume evolution curves are obtained as a function of time, and used to evaluate the end-diastolic volumes, end-systolic volumes, and ejection fraction. Results of the three cycles were averaged. The analysis was repeated a few weeks later to observe operator intravariability; new end-diastolic and end-systolic manual contours were traced by the same operator, for 10 phantom volumes. Different parameters were tested to ensure measurement reliability: variation of cardiac frequency (from 60 to 76 beats per s) with fixed volumes (50 and 120 mL for end-systolic and end-diastolic volumes, respectively), variation of end-systolic (from 50 to 130 mL) and end-diastolic (from 120 to 200 mL) volumes with fixed cardiac frequency (60 beats per s), influence of end-systolic instant determination (from -2 to $+2$ images) on volume reconstruction, influence of accuracy of calculus during computing (from 16 × 16 × 16 to a 64 × 64 × 64 reconstruction grid, leading to voxel sizes ranging from 4.4 to 1.1 mm for 50 mL volumes), and impact of probe rotational speed (varying from 4 to 8 rotations per s).

Fig. 7. Example of 8 successive phantom volumes reconstructed within a cardiac cycle with the LV 4-D method.

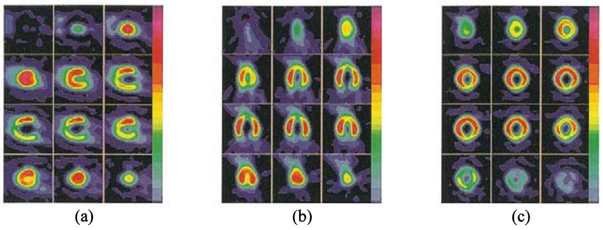

Fig. 8. Gated SPECT Images of the cardiac beating phantom: (a) frontal, (b) transverse and (c) cross-sections.

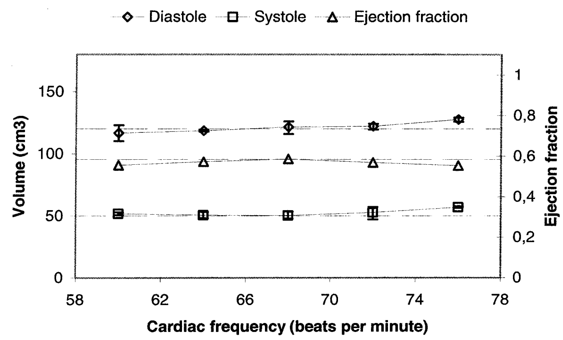

Fig. 9. Influence of the cardiac frequency on the LV 4-D measured volumes and ejection fractions. End-systolic and end-diastolic phantom volumes are, respectively, 50 and 120 mL, resulting in a 0.58 ejection fraction (horizontal dotted lines).

Gated SPECT

To carry out the different gated SPECT acquisition, 29 MBq of ^{99m}Tc were diluted in the volume (90 mL) comprising the myocardial wall. Images were acquired with a single-head DS7 gamma camera (GEMS, Buc, France), equipped with a high-resolution and low-energy collimator. Gating was done using a single period of 845 ms. The acquisition, using an energy window of 20% width—centered on 140 keV—produced 32 projections of 40 s over 180°. The 40-s time of each projection was chosen to approximate the statistical conditions commonly obtained with images of patients. Eight volumes gated to the electrocardiogram were acquired within each experiment, to comply with the standard clinical investigation conditions obtained with the described camera. Pixel size (6.90 mm) corresponding to the 64 $\times$ 64 matrix was calibrated according to local recommendations. The Mirage software, released by Segami (Ellicott City, MD), implements a filtered back projection, provides a complete sequence of 2-D images in three directions (Fig. 8), automatically reconstructs the observed volumes using a fifth order Butterworth filter, and then estimates the end-diastolic volume, end-systolic volume and ejection fraction.

Eight volumes were computed per cardiac cycle for each acquisition. To compare the isotope and US examinations, end-systolic (respectively end-diastolic) volume was increased from 50 to 130 mL (respectively 120 to 200 mL by 20-mL steps with a beat rate fixed to 60 per min.

RESULTS

This section starts by tests carried out to estimate the influence of different adjustable parameters on LV 4-D reconstructions. Then, it compares the LV 4-D and the gated SPECT results.

Heart rate variation

To test the influence of cardiac frequency on LV 4-D reconstructions, we increased the former from 60 to 76 beats per min (maximum frequency range available with the phantom), by steps of four. Each acquisition was carried out with an end-systolic volume of 50 mL, an end-diastolic volume of 120 mL (ejection faction of 0.58), a frame rate of 53 images per s, and a probe spin of 8 rotations per s. Figure 9 presents the results, and shows no significant discrepancy between each set of volumes.

Probe rotation speed variation

To check the influence of the rotation speed, we decreased this speed from 125 to 275 ms per rotation by steps of 50 ms per rotation with the frame rate; thus, varying the angle of successive images. All other parameters remained equal to those given in the previous paragraph, with a cardiac frequency adjusted at 60 beats per min. Figure 10 shows that, for a rotational speed between 125 and 175 ms per rotation, the measured volumes are constant and accurate. For a lower speed, a

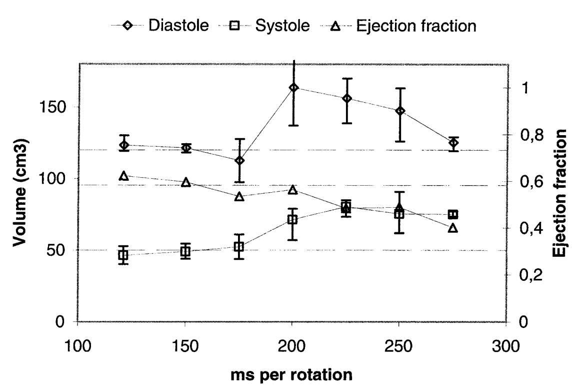

Fig. 10. Influence of probe rotational speed on measured volumes and ejection fractions. End-systolic and end-diastolic phantom volumes are, respectively, 50 and 120 mL, resulting in a 0.58 ejection fraction (horizontal dotted lines).

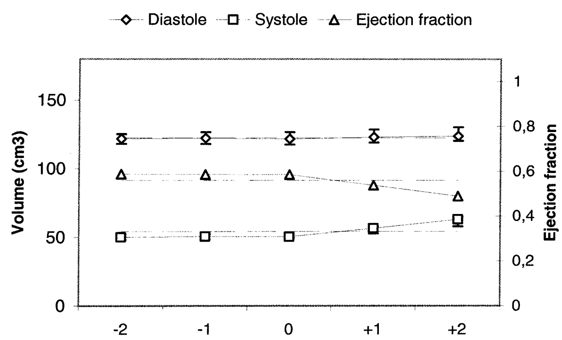

Fig. 11. Influence of determination of end-sytolic instant. End-systolic and end-diastolic phantom volumes are, respectively, 50 and 120 mL, resulting in a 0.58 ejection fraction (horizontal dotted lines).

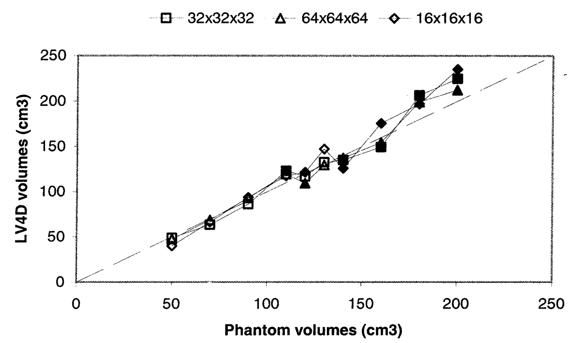

Fig. 12. Influence of the resolution on measured volumes (filled symbols for diastole, open symbols for systole).

significant divergence can be observed; two successive images are separated by an angle that does not allow three consecutive images to be used to reconstruct a volume correctly, especially in the case of a slight misalignment of the ultrasonic probe and the ventricle axis. Thus, the LV 4-D algorithm gives a noncoherent volume. This should be corrected by delineating more than three contours in diastole and systole, but would also lead to a time-consuming procedure.

Variation of the systolic instant determination

The first parameter the physician has to adjust when using the LV 4-D software is the systolic instant within the studied cardiac cycle. For this purpose, the cineloop can be played image by image to determine the smallest left ventricle, while each image is time marked in the ECG. However, this determination is sometimes difficult, because of the probe rotation leading to unusual cut planes. Figure 11 indicates volumes and ejection fraction variations when moving the end-systolic instant from 1 to 2 pictures more or less than the “true” systolic instant. Varying from − 2 pictures to + 2 pictures corresponds to an 80-ms variation, that is to say ± 10% of a standard cardiac cycle. All other parameters remained the same as in the previous paragraph. Figure 11 shows that advancing the contraction time (selection of the end-systolic instant before the “true” end-diastolic instant) has no significant influence on measured volumes. On the other hand, a delay of the systolic instant leads to an increase in the end-systolic volume. This is explained by the LV 4-D protocol. Three successive contours are drawn to reconstruct the end-systolic volume. The first one corresponds to the selected end-systolic instant. If the latter is delayed, the ventricle begins to relax quickly and cardiac volume increases significantly between the first and the third image.

Influence of volume accuracy

By default, the LV 4-D volumes are reconstructed using a 3-D grid of size 32 × 32 × 32 for interpolation. For physicians, this seems to be a good compromise between LV reconstruction accuracy and computing time. Nevertheless, it is possible to decrease the resolution to a 16 × 16 × 16 3-D grid, or increase it to 64 × 64 × 64. The greater is the grid size, the smoother is the visual aspect of the reconstructed volumes. However, the influence of the grid size had to be evaluated regarding to absolute volume measurements. Figure 12 shows volumes reconstructed using 16 × 16 × 16, 32 × 32 × 32 and 64 × 64 × 64 3-D grids. Respective computing times (PC computer Celeron, 1.0 GHz, 196 Mo RAM) are 2 s, 20 s and 2 min 40 s. The curves in Fig. 12 indicate that measured volumes are very close for all tested 3-D grids, and do not depend on the resolution. Because computing time increases drastically with the size of the grid, a higher resolution than 32 × 32 × 32 is not necessary for volume estimation, giving a reasonable balance between voxel size vs. computation time.

Intraoperator variability

This study showed that manual contour variability gives an error less than 7% of volume measurements reconstructed with LV 4-D. Thus, results seem very reliable for the use of US images.

Comparison of LV 4-D and gated SPECT volumes

Identical phantom parameters were adjusted for LV 4-D and gated SPECT examinations. Cardiac frequency was adjusted at 60 beats per min. Table 1 shows measured volumes for both techniques. Figures 13 and 14 show, respectively, volumes and ejection fraction measured by LV 4-D and gated SPECT, relative to the real phantom values. The results of both techniques are very similar, especially for ejection fraction, where the maximum difference between both examinations is less than

Table 1. Comparison of end-systolic (V_S), end-diastolic (V_D) volumes (in mL) and ejection fraction (EF) measured by LV 4-D and gated SPECT

Phantom values			LV 4-D			Gated SPECT		
V_S	V_D	EF	V_S	V_D	EF	V_S	V_D	EF
50	120	0.58	51.9	113.4	0.54	63	162	0.61
70	140	0.50	63.6	131.6	0.52	91	185	0.51
90	160	0.44	84.8	151.5	0.44	107	207	0.48
110	180	0.39	118.3	207.9	0.43	149	240	0.38
130	200	0.35	135.2	222.2	0.39	139	239	0.40

6%. Regression lines are adequate, and correlation coefficients are good: r = 0.997 for gated SPECT and r = 0.981 for LV 4-D. The comparison with the y = x line indicates a 33 % overestimation of measured volume by the gated SPECT examination, which leads to low values for high ejection fractions. Similarly, the correlation coefficient for ejection fraction is good; both methods give a correct evaluation of global systolic function on the phantom model.

DISCUSSION

Phantom

Many studies have been done on cardiac volume measurement to determine the accuracy and reliability with different modalities (Lowe et al. 1993; Reicheck 1987; Hoilund-Carlsen et al. 1984). They were tested mainly on animal or human models (Byrd et al. 1989) (ventricular mass and volume determined *post mortem*) and sometimes on static phantoms of different materials (Aakhus et al. 1994). Physiological cardiac rhythm and systolic function variability make studies difficult in living subjects. Similarly, static phantoms are very restrictive. Studies on isolated beating hearts have been carried out with US imaging (Smith et al. 1995), but cannot be realized with gated SPECT because the exam depends on

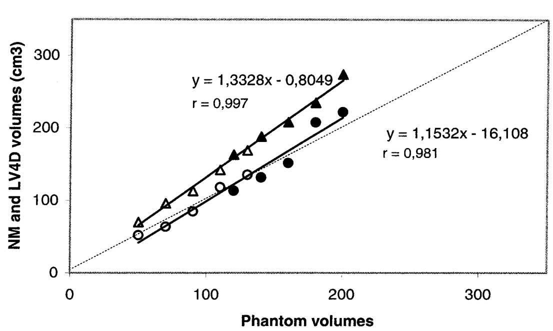

Fig. 13. Comparisons of the volumes (mL) of the phantom reconstructed from gated SPECT (systole = △, diastole = ▲) and LV 4-D (systole = ○, diastole = ●).

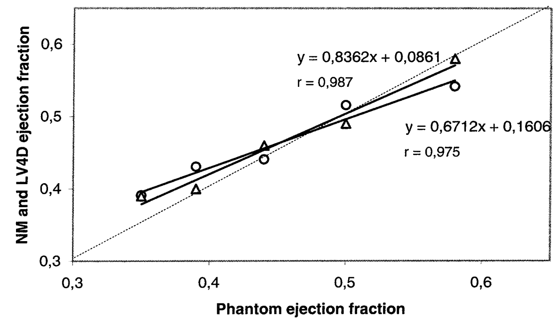

Fig. 14. Comparisons of the ejection fractions of the phantom reconstructed from gated SPECT (△) and LV 4-D (○).

radioisotope uptake by a living cell. A dynamic phantom is, therefore, the most suitable method to compare the new LV 4-D modality with scintigraphic examination before clinical evaluations. The range of volumes observed is wide (50 to 200 mL) and allows normal as well as pathologic situations to be simulated. On the other hand, the available frequency range is narrow on the phantom (only from 60 to 80 beats per min). It does not allow validation of both methods in the case of tachycardia or bradycardia.

4-D echography

The superiority of 3-D echocardiography over 2-D echocardiography to quantify ventricular volume has been demonstrated in animal models (Abdullah et al. 2000), digital models (Schroder et al. 1993), static phantoms (Rodevand et al. 1998) and clinical studies (Nosir et al. 1998). However, it has to be validated *in vitro* and *in vivo* cautiously, before any clinical practice. This has been done both in real-time 3-D echocardiography (static phantom) (Rusk et al. 2000) and in classic 3-D examination (using a slow step-by-step rotating probe). For example, recent work showed that a slight misalignment of a probe axis can lead to underestimations of the cardiac volumes, in 3-D echocardiography (Mannaerts et al. 2002).

Classic 3-D echocardiography is characterized by a long acquisition (up to 45 sequences of images) and computing time, even although several techniques shorten the duration of the examination (Kuhl et al. 2000; Tanabe et al. 2000). It is based on the superposition of different cardiac cycles acquired continuously, but at different instants. This implies a stable cardiac rhythm and a constant cardiac function during acquisition, which is not always the case.

LV 4-D reconstructs the ventricle volumes with data acquired within a single cardiac cycle. The limitations of 3-D echocardiography are, thus, not applicable. LV 4-D uses a standard US machine, unlike real-time

3-D imagers. A preliminary study carried out on healthy subjects showed good agreement between LV 4-D and 2-D echocardiography (Nguyen and Léger 2002). However, it was tested under "ideal" conditions: hearts of healthy subjects, leading to narrow ranges of cardiac volumes and heart rates. Consequently, the real influence of different algorithm parameters could not be intensively tested. A mechanical dynamic phantom was needed to carry out further validations.

LV 4-D intraoperator variability is equal to 6%, which is less than the 11% noted by Nadkarni et al. (2000) (3-D echography on porcine hearts). This result is fairly good because phantom echogeneicity is known to be suboptimal: the phantom was intended initially for calibration in nuclear imaging and not for echocardiography. The external Plexiglas shell is not removable and the wall geometry does not allow a perfect alignment of the US probe with the ventricular axis. Metallic masses at the bottom of the phantom perturb the accuracy of determination of the valve plane. Thus, the quality of imaging is worse than in human examinations, especially because second harmonic imaging does not notably improve image quality. Variability of absolute volumes from beat to beat can reach 15% in unfavorable cases. This is probably related to the manual drawing of the endocardial border, required for volume reconstruction, but carried out in suboptimal echogeneicity in the case of the phantom imaging. The influence of some other parameters, such as initial angulation of the probe (which changes from beat to beat), could not be assessed by this study, and requires numerical simulation. Nevertheless, this is not needed for the comparison of LV 4-D with gated SPECT.

LV 4-D volume and ejection fraction estimations present a good correlation with real phantom values, even under several parameter variations, which proves the robustness of the protocol. Increasing the reconstruction grid size (from $16 \times 16 \times 16$ to $64 \times 64 \times 64$) decreases the voxel size (respectively from 4.4 to 1.1 mm for 50 mL volumes), but does not alter absolute volume values significantly, even although it increases computing time. The $32 \times 32 \times 32$ resolution is chosen for default value because the quality of rendering is rough with the $16 \times 16 \times 16$ resolution. Heart beat rate (within the limits of the studied phantom) and end-systolic and end-diastolic volumes do not influence the reliability of measures. However, a slight overestimation is noticed on measured volumes in the case of "large" hearts. This is explained by the closeness of the internal and external walls (less than 5 mm) of "large" volumes, making discrimination between the two walls difficult. Nevertheless, the thinning observed with the simulated walls is quite unusual in human clinics, and real endocardial borders are much more visible than silicone ones. A correct selection of the end-systolic instant is crucial to produce relevant absolute volume measurements. In the case of indecision, picking one or two pictures earlier is more reliable because it reduces the error made on end-systolic volume and ejection fraction. Rotation speed is also an essential parameter of the LV 4-D protocol. Six contours (three in diastole, three in systole) drawn on consecutive images are required to estimate end-diastolic and end-systolic volumes, used, in turn, to determine the LV contours within each image available in the cardiac cycle. To ensure correct volume evaluations from these contours, the angle α between two successive images should be approximately equal to 60°. It depends on acquisition frame rate and probe rotation speed. In practice, RS is set to 8 rotations per s because *FR* is 53 images per s. Tests carried out showed that the optimal rotational speed may vary slightly without altering significantly the results. If *RS* is not reachable due to the *FR* value (probe rotation is limited to 8 rotations per s), manual drawing of all contours of the sequence is still possible, even although very time-consuming. This will be obviated when an automatic detection of contours not depending on estimated end-diastolic and end-systolic volumes is available.

LV 4-D vs. gated SPECT

Many studies have compared 3-D echocardiography with radioisotope imaging, principally with cavitary scintigraphy; Nosir et al. (1998) found similar results (with a maximum deviation of about 7%) in both examinations conducted on a set of patients. Likewise, real-time 3-D echocardiography showed a very good correlation with scintigraphic examination (Mondelli et al. 2001) both with animals and with patients (Takuma et al. 2001). These results were confirmed in the case of a pathologic heart (segmental hypokinesia) (Nosir et al. 1998). In those cases, 3-D echocardiography is reasonably better than classic echocardiography (biplane Simpson's method) to estimate ventricle volumes.

The comparison of LV 4-D and gated SPECT with measured phantom volumes showed a good agreement. The overestimation of volumes in scintigraphic examinations does not influence ejection fraction, and can be easily corrected by applying a proportionality coefficient (correction of the pixel size). Both techniques are rapid (15 min for LV 4-D, 40 min for gated SPECT), and meet the required conditions for further clinical evaluations.

Limitations

The constitution of the phantom provides accurate knowledge of the end-systolic and end-diastolic volumes, as well as beat rate. However, volume variation during the cardiac cycle is approximate because it is mechanically controlled (the shape of the camshaft that drives the piston of the pump) and weakly documented. Therefore, comparisons between the volume-time curves of the phantom and both studied methods are not possible.

Similarly, regional contraction of the cardiac phantom walls, theoretically homogeneous, depends on the local elastic properties of the silicone. The regularity of the contraction is, therefore, uncertain because the membranes are not intended for particular use in a phantom. An accurate local matching between the phantom volume and the LV 4-D and gated SPECT reconstructed volumes is, therefore, not possible.

CONCLUSION

4-D techniques are, theoretically, the best approaches to analyze the ventricle complexity with maximum accuracy. The results obtained in this study showed a straight correlation between real phantom volumes and volumes reconstructed with LV 4-D and gated SPECT. The agreement confirmed that both techniques can be used in conjunction for further human clinical evaluation. LV 4-D will be compared with gated SPECT, exploited as a reference examination. In the medium term, LV 4-D could improve echocardiography by emphasizing the quantization of volumes and reducing the learning process by physicians (Chuang et al. 1999), especially in stress echocardiography.

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