Three-Dimensional Echocardiography: In Vitro and In Vivo Validation of Left Ventricular Mass and Comparison With Conventional Echocardiographic Methods

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Objectives. This study aimed to validate a method for mass computation in vitro and in vivo and to compare it with conventional methods.

Background. Conventional echocardiographic methods of determining left ventricular mass are limited by assumptions of ventricular geometry and image plane positioning. To improve accuracy, we developed a three-dimensional echocardiographic method that uses nonparallel, nonintersecting short-axis planes and a polyhedral surface reconstruction algorithm for mass computation.

Methods. Eleven fixed hearts were imaged by three-dimensional echocardiography, and mass was determined in vitro by multiplying the myocardial volume by the density of each heart and comparing it with the true mass. Mass at diastole and systole by three-dimensional echocardiography and magnetic resonance imaging (MRI) was compared in vivo in 15 normal subjects. Ten subjects also underwent imaging by one- and two-dimensional echocardiography, and mass was determined by Penn convention, area-length, and truncated ellipsoid algorithms.

Results. In vitro results were r = 0.995, SEE 2.91 g, accuracy 3.47%. In vivo interobserver variability for systole and diastole was 16.7% to 27%, 14% to 18.1% and 6.3% to 12.8%, respectively, for one-, two-, and three-dimensional echocardiography and was 7.5% for MRI at end-diastole. The latter two agreed closely with regard to diastolic mass (r = 0.895, SEE 11.1 g) and systolic mass (r = 0.926, SEE 9.2 g). These results were significantly better than correlations between MRI and the Penn convention (r = 0.725, SEE 25.6 g for diastole; r = 0.788, SEE 28.7 g for systole), area-length (r = 0.694, SEE 24.2 g for diastole; r = 0.717, SEE 28.2 g for systole) and truncated ellipsoid algorithms (r = 0.687, SEE 21.8 g for diastole; r = 0.710, SEE 24.5 g for systole).

Conclusions. Image plane positioning guidance and elimination of geometric assumptions by three-dimensional echocardiography achieve high accuracy for left ventricular mass determination in vitro. It is associated with higher correlations and lower standard errors than conventional methods in vivo.

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Left ventricular mass is an important clinical variable to study in patients with various cardiac pathologies, including hypertension, valvular heart disease, heart failure and ventricular remodeling after myocardial infarction. The Framingham study (1) has demonstrated that elevated left ventricular mass is an independent and stronger predictor of morbid cardiac events and death than blood pressure or other risk factors, except for age. Conventional one-dimensional M-mode and two-dimensional echocardiographic methods of measuring mass are associated with comparatively high measurement variability, particularly in abnormally shaped ventricles (2). Alternate imaging modalities, such as magnetic resonance imaging (MRI) and computed tomography, are expensive, nonportable and not widely available for serial mass determination in patients. Therefore, a more accurate, reproducible echocardiographic method of computing cardiac mass would be useful, but to date none has been available.

Echocardiography is a low-cost, noninvasive, portable, easily repeatable modality useful for left ventricular mass determination. Current one-, (3,4) and two-dimensional echocardiographic methods (5–7) are limited by assumptions with regard to ventricular geometry and image position,
Three-dimensional echocardiographic method of left ventricular mass determination from unequally spaced, nonparallel, nonintersecting short-axis cross sections. **Top left,** First parasternal long-axis reference image. The vertical white lines are the lines of intersection of realtime guided short-axis cross sections positioned using the line of intersection display. **Top right,** Second parasternal long-axis reference image, which visualizes the apex. White lines represent the same guided short-axis images as those shown in the top left panel. **Bottom left,** Real-time short-axis image with the line of intersection of both reference images shown. **Bottom right,** Traced epicardial and endocardial boundaries of a digitized real-time short-axis image to be used for calculation of left ventricular mass by polyhedral surface reconstruction.

To overcome these limitations, we constructed a three-dimensional echocardiograph that registers transducer and image position and orientation in an independent, external three-dimensional spatial coordinate system (19). The problem of accurate image plane positioning has been solved by the novel use of an orthogonal reference image that intersects the real-time image, producing a "line of intersection" display (21) (Fig. 1). Use of this display enables the operator to more accurately position the real-time image planes to define a particular anatomic boundary or measurement shown in the reference image (22,23). To implement volume and mass computation, the ventricular endocardial and epicardial surfaces are reconstructed using a polyhedral surface reconstruction algorithm. The enclosed volumes are represented by a series of tetrahedrons, the volumes of which are computed and summed (24). The system has an overall volume computation accuracy of 0.4% in vitro (25). A comparison of its volume computation accuracy with that of MRI in vitro showed them to be comparable (26). A subsequent in vivo study of left ventricular volume computation by three-dimensional echocardiography also showed results comparable to MRI in normal human volunteers (27).

This report evaluates the results of our in vitro and in vivo validation studies using three-dimensional echocardiography for left ventricular mass computation. Multislice, multiphase spin-echo MRI using five cross-sectional images at enddiastole is used as a standard of comparison. This method has been validated independently for mass computation (28,29) and is a suitable standard of comparison, as demon-
strated by our previous in vitro (26) and in vivo (27) work. We also report a comparison of in vivo mass results by three-dimensional echocardiography with those obtained by one- (3,4) and two-dimensional (8) echocardiography.

**Methods**

Three-dimensional echocardiographic instrumentation. Three-dimensional echocardiography was performed with a real-time conventional two-dimensional echocardiograph (model 77020AC, Hewlett-Packard) linked to a three-dimensional acoustical spatial locator (model GP 8-3D, Science Accessories Corp.) that associates three-dimensional spatial coordinates with each image. The real-time images and spatial coordinates are transmitted to a personal computer (model AST Premium/286, AST Research) that controls system operation and provides means for subsequent analysis of the acquired three-dimensional data. The system, principles and methods of operation have been described elsewhere (21). Briefly, an array of three sound emitters is rigidly attached to a 2.5-MHz ultrasound transducer and are activated in sequence, sending 60-kHz sound waves to each of four overhead microphones mounted in a square array positioned 0.75 m above the patient. The time for the sound travel from each emitter to each microphone is measured, corrected for environmental conditions and used to calculate the X, Y, Z coordinates of the transducer and its image in a spatial coordinate system defined by the microphone array. The accuracy of the three-dimensional acoustic spatial locator is ~0.1% of the distance from the sound emitter to the microphone, permitting localization of each sound emitter to within 1 mm (30–32).

Three-dimensional echocardiographic “line of intersection display.” The “line of intersection display” is used to show the three-dimensional relation of several two-dimensional images. It allows the operator to see the position and orientation of the real-time image in relation to anatomic landmarks in its orthogonal, nonvisualized dimension. It is created by first saving in computer memory a long-axis reference image. When subsequent short-axis images approximately orthogonal to the long-axis image are acquired with their spatial coordinates, they intersect the long-axis reference image, defining a line common to both images, the line of intersection. This line is computed and displayed as a white line in each image (Fig. 1). As the real-time image is moved by the operator, the spatial coordinates are rapidly and repeatedly updated and the line of intersection is recomputed and redisplayed to show the changing relation of the real-time image to anatomic landmarks in the reference image (21). The ability to see this relation is helpful for guiding the position of the real-time image for linear measurements, for positioning end-planes at the aortic valve and apex and for optimally positioning image planes through the body of the ventricle to visualize endocardial boundaries. It is also useful for instructing sonographers and assessing their performance in positioning images (22).  

Three-dimensional echocardiographic image acquisition protocol. A series of images of a structure, such as the left ventricle, can be obtained that adequately depicts its size and shape. Subsequently, these images may be recalled from a computer file, their borders traced and the structure reconstructed by the computer. To establish a reference image and make use of the line of intersection display, images must be acquired in an appropriate sequence (Fig. 1). This sequence produces optimal positioning of the principal reference image, the parasternal long-axis image. Initially, a preliminary “positioning” short-axis image is acquired through the aortic valve. Next, a second positioning short-axis image is acquired close to the left ventricular apex. Then, the transducer is turned to obtain a real-time long-axis image. At this point, the line of intersection of the long-axis image with each positioning short-axis image is computed and displayed within all three images. These images are alternately viewed by pressing appropriate function keys on the computer keyboard. While viewing the positioning short-axis images, the operator adjusts the position of the long-axis image so that its line of intersection in each positioning short-axis image passes as closely as possible through the center of the aortic valve and the center of the left ventricular apex. When this result is achieved, the long-axis image has been optimally positioned and is saved in computer memory for subsequent use. Because the apex of the ventricle is usually not visible in a conventional parasternal long-axis view, two long-axis reference images are acquired; the second, at a lower interspace, includes the apex. The operator then uses these reference images, together with the line of intersection display, to selectively guide and position a series of registered, unequally spaced, nonparallel short-axis “data” images spanning the ventricle from the inferior surface of the aortic valve leaflets to the epicardial apex, as shown in Figure 1. The lines shown in Figure 1 represent the intersection of each acquired short-axis image with the reference long-axis images.

Three-dimensional echocardiographic polyhedral surface reconstruction algorithm. Epicardial and endocardial boundaries in each short-axis cross section are traced (Fig. 1), and the total volume subtended by each set of traced boundaries is determined using a polyhedral surface reconstruction algorithm (26). Each traced boundary is divided by interpolation into 180 equally spaced coordinate points, and the centroid of each boundary is automatically determined. Pairs of adjacent boundary coordinate points are then connected, creating a pair of triangular surface tiles for each pair of corresponding points. Each triangle is defined by two consecutive points on the same boundary and a single point on the adjacent boundary. Thus, the triangle is created by a single “contour element” and two “spans.” Boundary coordinate points are also connected to the centroids to define 180 separate wedges, or sectors. Each wedge is then decomposed into three separate tetrahedrons to yield a total of 540 tetrahedrons per slice. The volumes of the 540 tetrahedrons are summed to yield slice volume. Slice volumes are in turn
These, 15 subjects (7 men, 8 women; 23 to 41 years old; body surface area 1.38 to 2.17 m²) were enrolled for participation. One subject was excluded because of poor echocardiographic image quality and another because of poor magnetic resonance image quality. All subjects were voluntary participants and gave verbal informed consent before the study.

Subjects were imaged by three-dimensional echocardiography and gated cardiac MRI. Ten of these subjects (6 men, 4 women; 26 to 41 years old; body surface area 1.53 to 2.17 m²) also underwent left ventricular mass determination by conventional one- and two-dimensional echocardiography.

**Left ventricular mass by three-dimensional echocardiography.** Examinations were easily performed in the left lateral decubitus position and were of minimal technical difficulty. High quality images with easy and complete identification of all structures were available in all cases. Images were always acquired at end-expiration using the standardized three-dimensional echocardiographic imaging protocol described earlier using two reference long-axis images, and eight to nine short-axis slices were acquired using the line of intersection display. When significant patient motion occurred during image acquisition, the data set was discarded, and the procedure was repeated. Separate sets of reference images and short-axis images were collected at end-diastole and end-systole. Approximately 6 to 8 min was required for acquisition of each diastolic and systolic data set. Epicardial and endocardial boundaries were then manually traced for each image requiring ~1 min/boundary. Total volume subtended by the traced epicardial and endocardial boundaries was determined using the polyhedral surface reconstruction algorithm described earlier. Myocardial volume was then calculated by subtracting the endocardial volume from the epicardial volume. Myocardial mass was determined by multiplying this volume by 1.05 g/ml, the density of myocardium. Mass was determined from the end-diastolic images, as well as from the images with the smallest chamber dimension, or end-systolic images. Left ventricular mass was determined by two observers in a blinded manner, and the results were averaged.

**Left ventricular mass by one-dimensional echocardiography.** Subjects were imaged by unguided M-mode echocardiography, and left ventricular mass was obtained from a midventricular dimension using a regression-corrected, modified cube method or Penn convention (3).

**Left ventricular mass by two-dimensional echocardiography.** Left ventricular mass was determined by two-dimensional echocardiographic methods currently recommended by the American Society of Echocardiography on the basis of an area-length model and a truncated ellipsoid model (7,8). Both methods use the midventricular short-axis view at the level of the papillary muscle tips, generally the widest short-axis left ventricular diameter and apical four- and two-chamber views to maximize the length of the ventricle. The area-length method uses the entire major axis, whereas the truncated ellipsoid method divides the major axis at the widest minor axis into semimajor and truncated semimajor axes. End-diastolic frames were selected, digi-
tized and traced using a video display analysis system (Color Workstation, Freeland Medical System). Mass was calculated from the total area subtended by the epicardium and cavity area and their difference, or myocardial area (8). For tracing purposes, papillary muscles were considered part of the cavity.

**Left ventricular mass by MRI.** Magnetic resonance imaging was performed within 2 to 4 h of the three-dimensional echocardiographic study using a 1.5 T Sigma system (GE Medical). Subjects were connected to telemetered echocardiographic gating with respiratory compensation and then placed in a 30° left anterior oblique (left shoulder down) position placing the left ventricular long axis in the coronal plane. Initial “scout” proton imaging was obtained using a multislice, single-phase, gated spin echo sequence (repetition time one RR interval; echo time 20 s). The coronal images (long axis) were acquired in a 256 x 128 matrix. The scout image displaying the longest vertical axis of the heart was used to define the location for the subsequent slices used to calculate left ventricular volume. Five contiguous slices perpendicular to the long axis of the heart (short-axis views) were located on the scout image spanning the ventricle from base to apex. The slice thickness was 1 cm, but the interslice gap was adjusted so that the first slice transected the base of the heart and the fifth slice the apex. Images were then acquired using a gated, multiphase, multislice spin echo sequence with an echo time of 30 ms and repetition time equal to the RR interval. Images were collected into a 256 x 128-voxel matrix, using a total of two averages. Endocardial blood pool boundaries were enhanced by applying saturation outside the region of interest along the cephalocranial and mediolateral axis. End-diastolic images acquired 60 ms after the onset of the QRS complex were digitized into a video display analysis system (Color Workstation, Freeland Medical System). Total imaging time was 30 min/subject.

A summation of slices method was used to determine volumes defined by the endocardium and epicardium. Endocardial and epicardial borders of end-diastolic images were traced twice by two independent observers, and the corresponding area was multiplied by the slice thickness to obtain the slice volume. The interslice volume (gap volume) was estimated by averaging the area of the adjacent slices and multiplying this by the gap thickness. Endocardial volume was subtracted from epicardial volume to yield myocardial volume, which was multiplied by myocardial density 1.05 g/ml to yield myocardial mass. The results of both observers were averaged.

**Analysis.** Conventions for boundary tracing by one-, two-, and three-dimensional echocardiographic methods were as follows: 1) tracing was performed on the myocardial or white side of the black-and-white endocardial and epicardial boundaries; and 2) papillary muscles were excluded from the mass when they were not contiguous with the endocardium and were included in the mass when they were contiguous with the endocardial boundary. This convention was required for papillary muscle tracing because the polyhedral surface reconstruction algorithm does not accept discontinuous data. Boundary tracing of magnetic resonance images always excluded the papillary muscles from the myocardial volume. The mean mass values of two observers at end-diastole and end-systole were determined for one-, two- and three-dimensional echocardiography. End-diastolic mass by MRI was used for comparison.

For 15 normal subjects, interobserver variability for selection of images and boundary tracing was determined for three-dimensional echocardiography and MRI, as shown in equation 2, where $x_1$ = values of observer 1; $x_2$ = values of observer 2; and $n = 15$:

$$\frac{\sqrt{\sum (x_1 - x_2)^2/n}}{1/2 (\sum x_1/n + \sum x_2/n)}$$

Three-dimensional echocardiographic mass values obtained in systole and diastole were compared with MRI values using the Pearson correlation coefficient and simple regression. The $F$ test was used to test the null hypothesis that 1) MRI and three-dimensional echocardiographic diastolic measurements were identical, 2) MRI and three-dimensional echocardiographic systolic measurements were identical, and 3) three-dimensional echocardiographic diastolic and systolic measurements were identical, thus yielding the equation $y = x$. The regression estimates of the slope and intercept were simultaneously tested against the values of 1 and 0 (values for the line of identity) by the $F$ test (32).

**Results**

**In vitro validation.** The results of three-dimensional echocardiographic in vitro validation studies in fixed animal hearts are illustrated in Figure 2. Specimen masses ranged from 63.4 to 136.9 g. An excellent correlation between mass by three-dimensional echocardiography and true mass was obtained: $r = 0.995$, SEE 2.91 g, $p < 0.001$. The regression equation (Three-dimensional echocardiographic mass = 1.06(True mass) − 8.26) was not statistically different from the line of identity by the $F$ test. The root mean square percent error or accuracy was 3.47%.

**In vivo validation.** The results of three-dimensional echocardiographic in vivo validation studies using MRI as a standard of comparison are illustrated in Figure 3. Masses of the 15 normal subjects ranged from 70.2 to 149.5 g. Interobserver variability of three-dimensional echocardiography was 12.8% for diastolic values and 6.3% for systolic values. Interoobserver variability of MRI was 7.5% for diastolic values. Correlation coefficients and standard errors of the estimate between two observers for three-dimensional echocardiography were $r = 0.90$ and SEE 9.24 g for diastole and $r = 0.96$ and SEE 6.84 g for systole. The correlation between...
two observers for MRI measurements was \( r = 0.99 \) and SEE 4.07 g. Three-dimensional echocardiographic diastolic masses were not statistically different from three-dimensional echocardiographic systolic masses by the \( F \) test. The regression of left ventricular mass determined by three-dimensional echocardiography during diastole against MRI produced \( r = 0.895 \) and SEE 11.12 g. The regression equation for values obtained at diastole (Three-dimensional echocardiographic mass = 0.854[MRI mass] + 12.2) was not different from the line of identity by the \( F \) test. The regression of left ventricular mass determined by three-dimensional echocardiography during systole against MRI was \( r = 0.926 \) and SEE 9.24 g for the equation. Three-dimensional echocardiographic mass = 0.867(MRI mass) + 10.3. There was no significant difference between the latter regression equation and the line of identity by the \( F \) test using two degrees of freedom and \( n - 2 = 13 \).

**In vivo comparison of one-, two- and three-dimensional echocardiography with MRI.** Masses ranged from 74.8 to 149.5 g for the subset of 10 subjects who underwent comparison studies. Comparisons of the one-dimensional Penn convention and three-dimensional echocardiography with MRI are shown in Table 1. Interobserver variability by the Penn convention was 27.0% for diastolic values and 16.7% for systolic values. Compared with the Penn convention, three-dimensional echocardiography achieved a higher correlation coefficient (\( r = 0.725 \) vs. \( r = 0.882 \) for diastole; \( r = 0.788 \) vs. \( r = 0.908 \) for systole), lower standard error of the estimate (25.6 vs. 10.4 g for diastole; 28.7 vs. 10.8 g for systole) and a more significant \( p \) value (\( p = 0.018 \) vs. \( p = 0.001 \) for diastole; \( p = 0.007 \) vs. \( p = 0.001 \) for systole).

Comparisons of the two-dimensional echocardiographic area-length method recommended by the American Society of Echocardiography and three-dimensional echocardiography with MRI are shown in Table 1. Interobserver variability by the area-length method was 14.0% for diastolic values and 18.1% for systolic values. Compared with the area-length method, three-dimensional echocardiography achieved a higher correlation coefficient (\( r = 0.694 \) vs. \( r = 0.882 \) for diastole; \( r = 0.717 \) vs. \( r = 0.908 \) for systole), lower standard error of the estimate (24.2 vs. 10.4 g for diastole; 28.2 vs. 10.8 g for systole) and a more significant \( p \) value (\( p = 0.03 \) vs. \( p = 0.001 \) for diastole; \( p = 0.03 \) vs. \( p = 0.001 \) for systole).

The results of the comparison of the two-dimensional echocardiographic truncated-ellipsoid method recommended by the American Society of Echocardiography and the comparison of three-dimensional echocardiography with MRI are shown in Table 1. Interobserver variability by the truncated ellipsoid method was 16.3% for diastolic values and 16.6% for systolic values. Compared with the truncated ellipsoid method, three-dimensional echocardiography achieved a higher correlation coefficient (\( r = 21.8 \) g vs. \( r = 10.4 \) g for diastole; \( r = 24.5 \) g vs. \( r = 10.8 \) g for systole) and a more significant \( p \) value (\( p = 0.03 \) vs. \( p = 0.001 \) for diastole; \( p = 0.02 \) vs. \( p = 0.001 \) for systole).

**Discussion**

From these results, we conclude that three-dimensional echocardiography achieves high accuracy for left ventricular
mass determination in vitro. In addition, three-dimensional echocardiography also achieves highly accurate in vivo results in normal subjects. The standard error of the estimate for three-dimensional echocardiography compared with MRI in normal subjects is in the range 9 to 11 g. Improved interobserver variability is obtained by the three-dimensional echocardiographic method for values at end-systole compared with those obtained at end-diastole, probably as a result of superior endocardial boundary recognition. Left ventricular mass by three-dimensional echocardiography is also associated with higher correlation coefficients and at least a twofold improvement in the standard error of the estimate compared with conventional one- and two-dimensional echocardiographic methods. We believe that these results are achieved by using the line of intersection display for image positioning guidance and by the use of a polyhedral surface reconstruction algorithm that eliminates the need for geometric assumptions. No significant differences were noted between mass values obtained by three-dimensional echocardiography at end-diastole and those obtained at end-systole. This is consistent with the relative constancy of mass during the extremes of the cardiac cycle, as reported by other investigators (7,29,34,35).

**Comparison with conventional echocardiographic studies.** Previous echocardiographic methods of measuring left ventricular mass have used M-mode and two-dimensional techniques (2–7), all of which require geometric assumptions. Devereux and colleagues (3,4) described a one-dimensional echocardiographic technique, known as the Penn convention, that derives ventricular mass by a regression equation from an unguided unidimensional measurement of the thickness of the septum, posterior wall, and left ventricular chamber dimension. These investigators also showed, however, that the geometric assumptions required for M-mode methods cannot account for variability in left ventricular distribution of mass as the abnormal heart deviates from its normal shape. The observer variability associated with M-mode measurements made in our study is consistent with the values of 10% to 25% reported by the American Society of Echocardiography (35). The correlation coefficients obtained for one-dimensional echocardiography in our study are lower than those reported previously. This may be attributed to the relatively narrow range of mass values encountered in our normal subjects compared with the wide range of values reported in populations with heart disease. However, our standard errors of the estimate for one-dimensional echocardiography are in fact lower than those reported by previous investigators (6).

The American Society of Echocardiography (8) has recommended either the two-dimensional truncated ellipsoid or the area-length method of left ventricular mass computation, which use a single apical view and a midventricular short-axis view. These methods have been reported to be superior to one-dimensional echocardiographic methods, although they still use assumptions of ventricular shape and image relations (2.5). They require that the views be orthogonal with the ventricle and each other, a condition not verified and seldom attained (22). These assumptions are required because the position and orientation of each image is not spatially registered in conventional echocardiography, a major deficiency limiting quantitative capability. In addition, image plane positioning errors occur because the operator is unable to visualize anatomic landmarks in the dimension orthogonal to the real-time image plane. Both types of error lead to high measurement variability among operators (23) that limits test reliability for serial evaluation of left ventricular mass in individual patients. The correlation coefficients obtained for two-dimensional echocardiographic estimates of left ventricular mass in our study are somewhat lower than those reported in previous studies, in part because of the narrow range of mass values encountered in our normal subjects. However, our standard errors of the estimate were no greater than, and in fact were lower than, reported values of 31 to 71 g, depending on the algorithm used (5,6). The interobserver variability for mass computation by conventional two-dimensional echocardiographic methods is somewhat greater than those reported in some studies (36). This may be due to the conduct of the studies in laboratories specially dedicated to reporting quantitative two-dimensional echocardiographic measurements. However, there is a need for more improved accuracy and reproducibility in all laboratories before routine serial measurement of left ventricular volume and mass in individual subjects can be used clinically to provide information for diagnosis, management and assessment of therapy.
Comparison with previous three-dimensional echocardiographic studies. The results of our in vitro validation of mass are consistent with our previous in vitro validations of volume determination using various phantoms, a stylized pin model, nonuniform latex balloon phantoms and glutaraldehyde sheep and pig hearts. Low standard errors and accuracy ranging from 1.6% to 2.3% were obtained in these studies (25,26).

Previous in vivo experience with three-dimensional echocardiography for ventricular mass determination has been limited only to preliminary results in transplant patients reported by McGaughey et al. (20). However, previous in vivo validation of the three-dimensional echocardiographic system and the polyhedral surface reconstruction algorithm has been carried out by us for left ventricular end-diastolic and end-systolic volumes using MRI as a clinical standard of comparison. Correlations for end-diastolic volume \((r = 0.92, \text{SEE} 6.99 \text{ml})\) and end-systolic volume \((r = 0.81, \text{SEE} 4.01 \text{ml})\) by three-dimensional echocardiography were significantly improved compared with the conventional two-dimensional echocardiographic apical biplane summation of disks method \((r = 0.48, \text{SEE} 20.5 \text{ml for diastole}; r = 0.70, \text{SEE} 5.6 \text{ml for systole})\) (31). Thus, three-dimensional echocardiography reduces left ventricular volume computation error by one-half to two-thirds by the elimination of geometric assumptions and by improving image plane positioning through use of the line of intersection display. The current study achieves comparable results for left ventricular mass and improvement of three-dimensional echocardiography over one- and two-dimensional echocardiography.

Advantages of the line of intersection display. Our three-dimensional echocardiographic method overcomes two major limitations of conventional echocardiography and can therefore be expected to achieve better measurement of left ventricular mass. In addition to measuring between image planes, and thus eliminating the need for geometric or spatial assumptions, our system improves image positioning. To reduce or eliminate interobserver variability, two independent operators not only must use the same standardized procedure for measuring a dimension but also must reproduce the same tomographic image plane. High interobserver measurement variability of left ventricular dimensions was noted in previous two-dimensional echocardiographic studies because of variable positioning of the tomographic image plane (22,23). Good measurement reproducibility requires reduced variability in the position and angulation of the ultrasound image in its orthogonal, nonvisualized dimension. This reduction can now be achieved with a three-dimensional echocardiographic system capable of orienting and guiding positioning of the interrogating beam. This feature is provided on-line in an interactive manner and provides immediate three-dimensional spatial information to the operator to help direct the course of an examination (21). Other three-dimensional echocardiographic systems have not provided this feature. By observing the position, angulation and changing relations of the line of intersection to anatomic landmarks in the orthogonal reference image, the operator can guide the movement of the real-time image in the nonvisualized dimension. This corrects undesirable displacements, angulation or rotation of the image and may improve examination quality by correcting positioning errors.

When applied to the measurement of ventricular mass, line of intersection guidance is used to accurately position the short-axis end-planes at the base and apex. It is also used to achieve accurate angulation and spacing of images through the body of the ventricle to ensure accurate representation of the curvature of its walls. As a result of the polyhedral surface reconstruction algorithm and line of intersection image positioning guidance, three-dimensional echocardiography is more flexible in positioning images than one- or two-dimensional echocardiographic methods and advantageously uses additional echocardiographic windows.

Advantages of polyhedral surface reconstruction algorithm. Use of the polyhedral surface reconstruction algorithm eliminates the need to make any geometric assumptions about ventricular size or shape. It also permits acquisition of unevenly spaced, nonparallel images to define the ventricular surface. This flexibility allows use of multiple cardiac windows to optimize representation of the ventricle rather than restricting imaging to a predefined set of views acquired from one window. It is generally easy to obtain a set of short-axis images adequately depicting the entire left ventricle when multiple windows and nonparallel, unevenly spaced images can be used. When geometric constraints or assumptions are required, such as obtaining perpendicular images, it is often difficult or impossible to achieve satisfactory endocardial boundary definition. Because nonparallel cross sections can be used by the three-dimensional algorithm, image planes can be flexibly positioned so that endocardial and epicardial boundaries are optimally visualized.

Sources of error and study limitations. In our previous in vitro investigations (25,26), our three-dimensional echocardiographic system proved to be highly accurate and did not introduce new measurement errors. The accuracy of the locator is \(-0.1\%\) of the measured distances (30–32). The system accuracy in measuring volumes in vitro is \(-0.4\%\) (25). A limitation of this study is that the in vivo results reported in this study involve only small numbers of normal subjects and thus must be viewed as preliminary. Another limitation is that one- and two-dimensional echocardiographic studies were not performed on the same day as the three-dimensional echocardiographic and MRI studies. Further validation is required for mass determination in large numbers of patients with abnormal ventricles.

The method of data acquisition used in this study is to place and hold the imaging transducer in a desired location, acquire a single set of spatial coordinate data and subsequently acquire a series of 16 video images over the next second. Hence, a single spatial data set is assigned to each image. The alternative method is to continuously acquire spatial coordinate data during image acquisition, assigning a
new set of spatial data to each video image as it is acquired (9). Both methods, in the absence of respiratory gating and registration of body motion, are subject to misregistration of data. If respiratory or whole-body motion occurs during data acquisition, subsequent data will be misregistered with respect to data acquired before the motion, regardless of whether the method of data acquisition is discrete or continuous. Our discrete method of data acquisition is also subject to potential error because of transducer motion occurring during the 1-s acquisition of 16 video frames. During acquisition, two ventricular systoles and the diastolic interval between them are captured when heart rates are normal. To ensure that transducer motion does not introduce significant error, we demonstrated no significant difference in computed ventricular volumes using the first end-diastolic image of each data set compared with the second end-diastolic image of the data set (27). From these results, we infer that although transducer motion during image acquisition by the discrete method is theoretically possible, in practice it does not introduce significant error into the results of mass computation.

The polyhedral surface reconstruction algorithm slightly underestimates ventricular volume. The algorithm approximates the ventricular surface by a series of short straight chords that are connected to form triangles. The chords between points on the traced boundaries (contour elements) are very short and accurately approximate the curve. Those between traced boundaries (spans) are variably longer, depending on the spacing chosen by the operator. Because the curves being traced are usually convex, there will be a slight tendency for the straight line approximation underlying the surface to be shorter than the surface. As a consequence, the volume lying between the true surface and the surface approximated by the triangles will be omitted, resulting in a slight underestimation of volume. However, when 7 to 10 cross sections are used to represent the adult ventricle, based on the work of Weiss et al. (37), significant underestimation is not introduced.

Boundary tracing error remains a significant potential source of error with all imaging methods and by far outweighs other errors mentioned previously. However, satisfactory accuracy and reproducibility can be achieved by careful adherence to a well-defined rule for tracing. In our experience, tracing endocardial and epicardial boundaries on the "white" side of the black and white boundary has produced the best results. Another source of error is difficult visualization of the tip of the ventricular apex. Frequently, the apex will underlie a rib, making it difficult to image regardless of transducer position. Use of the second parasternal long-axis reference image over the apex greatly helps its accurate localization, however. Variable inclusion of the papillary muscle as part of the myocardium introduces another source of error to ventricular mass computation.

Magnetic resonance imaging has been validated in vivo for left ventricular mass determination in animal studies (28,29). However, formal in vivo human anatomic validation has not been performed. Magnetic resonance imaging is known to slightly overestimate myocardial mass because of partial volume effects and improper border definition (28). Partial volume effects are most important in areas of the heart that exhibit significant curvature of the epicardium or endocardium, particularly at the apex. Improper positioning of the patient may result in partial volume errors arising from oblique cross sections. Five 1-cm thick slices were obtained in accordance with previous in vivo MRI validation studies (28,29). Use of additional slices may have reduced partial volume effects but would have significantly increased the image acquisition time. Border definition remains a source of error of all imaging techniques including MRI. However, saturation outside the region of interest was applied to enhance endocardial-blood pool boundaries. Mass at end-systole was not used in the analysis because of the potential underestimation of mass associated with base to apex translation of the heart during systole. However, this effect is small, and previous investigators have reported no significant differences among values obtained at end-diastole and at end-systole (28).

Conclusions. The results of this study demonstrate that three-dimensional echocardiography using the polyhedral surface reconstruction algorithm and the line of intersection display for image positioning guidance provides an accurate and reliable estimate of left ventricular mass comparable to MRI in normal subjects. Furthermore, three-dimensional echocardiography is superior in accuracy and interobserver variability compared with conventional one- and two-dimensional echocardiographic methods for mass determination in normal subjects.

Implications. To date, serial quantitation of left ventricular mass has remained primarily a research tool, not a clinical one, in part because convenient, accurate and reproducible methods to measure it have not been developed. The availability of an accurate reproducible line of intersection-guided three-dimensional echocardiographic method of estimating left ventricular mass may have great benefit for routine clinical use as well as for large-scale clinical studies. We anticipate that the variance associated with three-dimensional echocardiographic measurements will be sufficiently small to allow it to be relied on for serial determinations in individual subjects. However, good performance in normal subjects does not ensure good performance in distorted hearts. To establish its serial use in patients, further validation studies with large numbers of patients with abnormal ventricles are required. The improvement in three-dimensional echocardiography over one- and two-dimensional echocardiography is expected to be further enhanced in this group of patients who have ventricular abnormalities that make geometric and image plane positioning assumptions more tenuous. Inasmuch as echocardiography is low in cost, noninvasive, portable, easily repeatable and widely available, this new three-dimensional capability for quantitative measurement of left ventricular mass shows promise.
for serial evaluation of patients with hypertension, heart failure and myocardial infarction.

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