Three-dimensional Echocardiography

In Vivo Validation for Right Ventricular Volume and Function

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Background Current two-dimensional echocardiographic measures of right ventricular volume are limited by the asymmetrical and crescentic shape of the ventricle and by difficulty in obtaining standardized views. Three-dimensional echocardiographic reconstruction, which does not require geometric assumptions or standardized views, may therefore have potential advantages for determining right ventricular volume. Three-dimensional techniques, however, have not been applied to the right ventricle in vivo, where cardiac motion and contraction could affect accuracy. The purpose of this study was to determine the feasibility and accuracy of three-dimensional echocardiographic reconstruction for quantifying right ventricular volume and function in vivo. In particular, it was designed to test the accuracy of a newly developed system that provides rapid, efficient, and automated three-dimensional data collection (minimizing motion effects) and takes advantage of the full three-dimensional data set to obtain volume.

Methods and Results The three-dimensional system was applied to reconstruct the right ventricle and measure its volume and function during 20 hemodynamic stages created in five dogs. Actual instantaneous volumes were measured continuously by an intracavitary balloon connected to an external column. Hemodynamics were varied by volume loading and induction of ischemia. Three-dimensional reconstruction successfully reproduced right ventricular volume compared with actual values at end diastole (\(y=1.0x-3.4, r=0.99, \text{SEE}=1.8\) mL) and end systole (\(y=1.0x+2.0, r=0.98, \text{SEE}=2.5\) mL). The mean difference between calculated and actual volumes throughout the cycle was 2.1 mL, or 4.9% of the mean. Ejection fraction also correlated well with actual values (\(y=0.96x-0.3, r=0.98, \text{SEE}=3.3\%\)).

Conclusions Despite the irregular crescentic shape of the right ventricle, this newly developed three-dimensional system and surface algorithm can accurately reconstruct its shape and quantitate its volume and function in vivo without geometric assumptions. The increased efficiency of the system should increase applicability to issues of clinical and research interest. (Circulation. 1994;89:2342-2350.)

Key Words • right ventricle • volume • echocardiography

Since the introduction of echocardiography, there has been long-standing clinical and research interest in assessing right ventricular (RV) dimensions, area, and volume by M-mode and two-dimensional echocardiography.\(^\text{1-17}\) The M-mode dimensions obtained from the parasternal long-axis view, however, are subject to variation with RV position relative to the chest wall and with patient position.\(^\text{18}\) Two-dimensional echocardiographic RV volume determinations, whether based on correlations with two-dimensional dimensions,\(^\text{2,11}\) angiographic area-length formulas,\(^\text{3,5,8,10,12,13}\) or Simpson's rule methods,\(^\text{5,7,9}\) have been limited by the complex and crescentic shape of the RV. Correlations between two-dimensional echocardiographic volumes and those obtained by angiographic or cast studies have been variable because the views selected often do not include the infundibulum\(^\text{6,7,8,12,14}\) and because of difficulties in obtaining the set of two perpendicular views used in angiography\(^\text{9,22}\) and imaging them in a standardized fashion.

Three-dimensional echocardiography, which has recently been validated for determining left ventricular (LV) volume,\(^\text{23-39}\) has potential advantages for determining RV volume because it reconstructs multiple two-dimensional images, eliminating the need for geometric assumptions and individual standardized views. Although it has been tested for determining the volume of RV casts and pressure-expanded specimens in vitro\(^\text{40,41}\) with promising results, it has never been validated in vivo with cardiac motion or contraction or with RV volume loading or ischemia.

Therefore, the purpose of this study was to determine the feasibility and reliability of three-dimensional echocardiography for quantifying RV volume and function in vivo using a canine model in which instantaneous RV volume can be measured directly with an intracavitary balloon connected to an external column to provide an ideal standard for volume measurement. The three-dimensional system used in this study has recently been developed to provide rapid, efficient three-dimensional data collection and take advantage of the full three-dimensional data set for surfacing and volume calculation.\(^\text{39}\) Although this system has been validated for the LV in vivo,\(^\text{39}\) its ability to reconstruct the more complex, narrow, and irregular RV in vivo remains to be proved.

Methods

In Vivo Model

Five mongrel dogs with a mean weight of 28.4±2 kg were anesthetized with pentobarbital (30 to 50 mg/kg IV), intu-
The resulting RV volumes ranged from 10.2 to 72.9 mL in two dogs, RV ischemia was produced by right coronary artery ligation to represent a full range of ejection fractions (18% to 77%). During each of the 20 hemodynamic stages, three-dimensional echocardiographic imaging was recorded simultaneously with video recording of the fluid column height. The studies conformed to the guiding principles of the American Physiological Society and received institutional approval.

**Three-dimensional Echocardiography**

**Data Acquisition**

The heart was scanned through a water bath standoff with the 3.5-MHz transducer of a Hewlett-Packard phased-array sector scanner (77020A). The RV was scanned in an intersecting series of short-axis sweeps (apex to infundibulum) and long-axis views from an anterior or a para-apical window (including long-axis views of the RV inflow and outflow tracts as well as a view in the four-chamber plane containing the RV apex). The three-dimensional positions of the images were recorded automatically and in real time by using three spark gap–locating devices attached to the transducer. These devices were placed on a plate perpendicular to the long axis of the transducer; the plate was mounted on a Pexiglas sleeve reproducibly fixed to the transducer. During the scan, the three spark gaps were fired in rapid succession by a microprocessor (Science Accessories, Inc); a square array of four microphones continuously received and timed the arrival of sound emitted from the spark gaps and therefore could locate each one by triangulation. Both the ultrasound machine and the transducer-locating system were interfaced to a single 386-series personal computer (SUN 386, Sun Microsystems), which encoded the positional data in real time as a binary pattern that was overlaid on an unused portion of the imaging video signal. The composite was then recorded on videotape (Fig 2), so that each video frame had the data required for three-dimensional reconstruction without need for manual coordination; scan time and storage requirements for this system are therefore the same as for the corresponding two-dimensional scans.

**Data Analysis**

After data acquisition, the same computer system and video board (AT Vista, True Vision) were used for data retrieval. For each RV reconstruction, 13 or 14 different tomographic images were selected from video playback to span the entire ventricle (inflow, outflow, and sinus portions) in intersecting planes. The selected images were then digitized, and the locating data were automatically decoded within 1 second to calculate the three-dimensional location of each image. End-diastolic images were defined as those with the largest RV cavity area (closest to and just following the QRS peak deflection on the simultaneous ECG), and end-systolic images were defined as those with the smallest RV cavity area (therefore corresponding to the greatest height of fluid in the column). To avoid mixing images in different spatial positions due to heart translation with respiration, only images in the quiet end-expiratory phase were used, as determined by a respiratory gating signal recorded on the videotape. The cavity borders in the selected images were then traced using a digital tracing device (Summagraphics Inc), with end diastole traced in red and end systole in green, to allow separate reconstructions and volume calculations for those time points. Areas of lateral dropout or indistinct borders were not traced as the computer algorithm was designed to tolerate incomplete or partial traces. During and after tracing, the group of traces could be displayed by the computer in three-dimensional space, so that the observer could review the consistency of tracing and the adequacy of sampling. Endocardial borders were traced up to the junction of the column and the RV infundibulum and outflow tract, to include the entire volume contained within the balloon.
The surface of the RV was then reconstructed, and its volume was calculated using a surfacing algorithm that takes advantage of the full three-dimensional data set. In brief, an initially spherical template was used to create an array of 800 latitude and longitude grid points. Rays were then drawn from the center of the sphere through each grid point, and the length of each ray was calculated to provide the best weighted fit to the actual traced borders in its vicinity. Any missing data points were filled in using a weighted fit to interpolate between nearest neighbors based on distance between grid points. The ends of the rays were connected to form a surface. Ventricular volume was obtained by summing the volumes of tetrahedrons formed by connecting the surface points to the center. The end-diastolic and end-systolic RV volumes could be obtained rapidly by the computer by selecting which color traces to enter into the surfacing algorithm. Stroke volume and ejection fraction were calculated from these volumes in a standard manner.

Statistical Analysis

Results for RV end-diastolic, end-systolic, and stroke volumes as well as ejection fraction calculated by three-dimensional echocardiography were compared with actual values by linear regression analysis. Ninety-five percent prediction limits were calculated with the RS1 statistical package (Bolt, Beranek and Newman, Inc). The mean difference between three-dimensional and actual values was also calculated. The error (three-dimensional minus actual volume) was analyzed as a function of actual volume for systolic and diastolic stages. To determine whether linkage of results in different animals affected results, multiple linear regression analysis of three-dimensional versus actual volumes was performed (RS1) with animal number and animal number times actual volume (interaction term) as additional independent variables. Interobserver variability in border tracing was expressed as the standard deviation of the differences between the measurements of two observers who independently traced and reconstructed 10 ventricles from videotaped images. Intraobserver variability in border tracing was similarly determined by one observer repeating the measurements for 10 ventricles.

Results

Reconstructed Images

Fig 3A is an example of the reconstructed traces showing the RV apex, outflow tract, inflow region, and curving septal surface. The corresponding surfaces used for volume calculation are shown in Fig 3B and 3C.

RV Volume

RV end-diastolic volumes calculated by three-dimensional reconstruction agreed well with actual values ($y=1.03x-3.4$, $r=.99$, SEE=1.8 mL; Fig 4A, top, and Table), with a small degree of underestimation: the mean difference between three-dimensional and actual values was $-1.6\pm1.7$ mL ($P<.05$; Fig 4A, bottom). Agreement was similarly good for RV end-systolic volume ($y=1.01x+2.0$, $r=.98$, SEE=2.5 mL; Fig 4B, top, and Table), with a small degree of overestimation: the mean difference between three-dimensional and actual values was $1.9\pm2.4$ mL ($P<.05$; Fig 4B, bottom).
Fig 3. A, Reconstructed traced borders of a beating right ventricle, with inlet region at the upper left, apex below, and outlet at the upper right. Diastolic traces (in red) and systolic traces (in green) are shown together (left) and separated for a stage with decreased right ventricular systolic function. B, Diastolic traces (left) combined with the corresponding surface used for volume calculation (right). C, Systolic traces (left) combined with the corresponding surface used for volume calculation (right).
mean difference between all three-dimensional and actual values of RV volumes was 2.1 mL, or 4.9% of the mean. There was no significant relation between volume errors and actual volume by linear regression (Fig 4A and 4B, bottom: for diastole, \( y=0.03x-3.4, r=.19, P>.04 \); for systole, \( y=0.002x+2.0, r=.04, P>.95 \). Multiple linear regression analysis showed no significant effect of animal number on the relation between three-dimensional and actual volumes in either systole or diastole and no significant effect of the interaction (animal number multiplied by actual volume; \( P>.09 \) for diastole, \( P>.6 \) for systole).

**RV Function**

RV stroke volume correlated well with actual values (\( y=0.92x-1.1 \)), with a correlation coefficient of .97 and a standard error of 2.5 mL (Fig 4C, top, and Table). The mean difference between three-dimensional and actual values was \(-3.5 \pm 2.6 \) mL (\( P<.05 \); Fig 4C, bottom). Correlation was similarly good for RV ejection fraction (\( y=0.96x-0.03, r=.98, \text{SEE}=3.3\% \) (Fig 4D, top, and Table). The mean difference between three-dimensional and actual values was \(-4.5 \pm 3.3\% \) (\( P<.05 \); Fig 4D, bottom).

**Observer Variability**

The interobserver variability in border tracing of the three-dimensional method was 1.86 mL, or 4.0% of the mean (Fig 5). The corresponding intraobserver variability was 1.23 mL, or 2.6% of the mean.

**Discussion**

The results of this study demonstrate that the three-dimensional echocardiographic system described can accurately reconstruct the RV and assess its volume and function in vivo compared with a directly measured standard. It uses all of the collected three-dimensional data for the volume calculation, without the need for geometric assumptions or standardized two-dimensional views.

**Limitations of Current M-Mode and Two-dimensional Methods**

The complexity of RV shape has limited attempts to calculate its volume from a single dimension or an area.
measurement.\textsuperscript{1-4,15-18} Even biplane approaches\textsuperscript{5-14} based on angiographic studies\textsuperscript{19-22} have produced only variable correlations and agreements with angiographic or radionuclide data because of several potential factors: (1) limitations of simplified geometric formulas in describing the complex RV, which is crescentic and asymmetrical, with a separate infundibulum; (2) difficulty in obtaining the two standardized orthogonal views with a common long axis, which are required for application of both Simpson's rule and biplane area-length methods; and (3) exclusion from geometric models of the RV outflow tract or infundibulum, which may account for 25% of the total RV volume.\textsuperscript{4,14}

Three-dimensional Echocardiography

Three-dimensional reconstruction overcomes the above limitations by reconstructing the ventricle without the need for simplifying geometric assumptions or standardized imaging planes. Its accuracy for assessing the more symmetrical left ventricle has been previously demonstrated both in vivo and in vitro.\textsuperscript{23-33,35-39} For the more irregular RV, however, its potential has until now only been assessed in vitro.\textsuperscript{40,41} The present study further validates its ability to reconstruct the RV and determine its volume and function in vivo, in the presence of cardiac motion, contraction, volume loading, and ischemia.

Three-dimensional reconstruction, in both the present study and a previous study,\textsuperscript{40} provides several other advantages.\textsuperscript{39} In brief, it combines multiple intersecting planes, improving the consistency of border detection by allowing each traced image to be reviewed in three-dimensional relation to the others; in particular, the current system allows such checking immediately during the tracing process. Also, the surfacing algorithm has the ability to accept partial traces, with missing data being filled in from intersecting views or by the surfacing algorithm itself; this can be especially important for the RV, parts of which may be difficult to visualize in a given view. The averaging effect of the surfacing algorithm will also minimize the impact of isolated tracing errors. In addition, the spark gap-locating system permits the operator to vary transducer position and scan plane to optimize image quality. Last, the three-dimensional system provides a surface that can be viewed and rotated to improve three-dimensional appreciation. The current system also provides several additional advantages. The location data are generated in real time and recorded simultaneously and automatically with the two-dimensional images, eliminating the need for manual coordination. Also, rapid data collection minimizes potential errors due to subject motion or respiration and ensures that positions and images are obtained at the same time. The surfacing algorithm uses the full strength of the intersecting three-dimensional data set to produce a polyhedral-type volumetric calculation.\textsuperscript{38,44} It therefore avoids the need for empirical correlation to estimate
Three-dimensional Echocardiographic Volumes vs Actual Volumes

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EF indicates ejection fraction; 3D, three-dimensional. Values are in milliliters.

areas in multiple parallel two-dimensional slices⁴⁵ (although that was not the central feature of the previous method or the major contributor to its accuracy⁴⁶).

Limitations and Future Work

There are several sources of variability in this method, including (1) the resolution of spark gap location (<1 mm by selection of data sets to have the computed distances between spark gaps differ by <1 mm from actual values); (2) observer variability; (3) a tendency of the surfacing algorithm to round out sharply protruding edges, such as the narrow apex and the distal outflow tract, although total volume tends to be preserved by the averaging procedure used to calculate the grid points; and (4) in the most narrow, crescentic end-systolic ventricles, a tendency of the surfacing algorithm to create slight localized bulging of the calculated surface because of the way it fills in grid points between actual traced images (it averages adjacent ray lengths as radii for a spherical template). These last two effects are likely causes of slight underestimations of end-diastolic volume and overestimations of end-systolic volume (Table); these errors were independent of chamber size and thus primarily affect small-chamber volumes. The systolic overestimation in particular may be less of a problem when volume estimation is most important clinically, for example, in patients with RV pressure or volume overload, in whom the cavity will be broader, decreasing the tendency of the algorithm to produce localized bulges. Ideally, errors may be reduced by projecting grid points from more than one focus to maximize the ability of the surfacing algorithm to adapt to local geometry. Nevertheless, the results of this study indicate that these effects are acceptably small for the ventricles examined. Greater variability, however, is likely in the clinical setting, related to a variety of factors. Respiration can cause changes both in the position of the heart (relative to the external frame of reference) and in its size and shape. Such variability should decrease with respiratory gating, whereas variability of cycle length could be dealt with by selecting beats within a specified range of cycle length for reconstruction. Variability caused by patient motion can be minimized by rapid acquisition as provided by this

Fig 5. Bar graph of interobserver variability. The height of the paired bars represents the volumes of each of 10 individual ventricles reconstructed by two independent observers. RV indicates right ventricle.
system; acquisition could be made even faster by transducers providing two simultaneous orthogonal views or multiple views by phased-array parallel processing. Limited image quality and acoustic access will decrease the number of planes available for reconstruction. The ability to reconstruct views of the RV apex along with a parasternal sweep while maintaining an acoustic line of sight for spark gap localization may require special attention: (1) short-axis views can be swept until the apex is reached, missing only on apical tip; (2) preferably, the patient can be oriented so that the long axis of the RV lies at an angle of 30° to 45° to the microphone array and both parasternal short-axis and par- apical long-axis views can be reconstructed (an approach applied to the left ventricle [Reference 39, Fig 8] that may be easier to apply to the more anterior and medial RV apex); and (3) in some patients, a view in the four-chamber plane can also be obtained anteriorly.

Regarding the model used, it must be emphasized that the purpose of this study was not simply to validate RV stroke volume but primarily to prove the accuracy of the three-dimensional method for calculating actual RV volumes against an ideal standard. The intracavitary model, as developed by Suga and Sagawa and modified by Weiss et al and the authors of the present study, serves this purpose. As implemented, even a 3-mm error in column height, which could be read to the nearest millimeter, produced less than 1 mL error in volume. One potential concern is the effect of RV trabeculations on the ability of any method to measure volume. This is less of a concern for three-dimensional than for other methods, however, because the averaging effect of the surfacing algorithm will tend to minimize potential inaccuracies caused by localized trabeculations in individual two-dimensional images.

With the improved efficiency of three-dimensional data acquisition provided by this system, endocardial border definition has become the most time-consuming step, requiring 10 to 15 minutes depending on observer experience, the number of images (13 or 14 in the present study), and their complexity. This time could be reduced by defining the minimal number of views required and by using new methods to automate or semiautomate border extraction based on signal amplitude or flow. Such systems could be particularly strong when applied to a three-dimensional data set because gaps in individual two-dimensional images could not be filled in from other images using minimal-cost functions that optimize the detection of a spatial border.

Summary
Despite the irregular crescentic shape of the RV, this newly developed three-dimensional system can accurately reconstruct its volume and assess its function in vivo without geometric assumptions or the need for standardized two-dimensional planes. The increased efficiency of this system, which allows rapid three-dimensional data acquisition, has the potential for increasing applications to concerns of clinical and research interest; these validation studies lay the foundation for subsequent work.

Acknowledgments
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