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=.99, SEE=1.8 mL) and end systole (y=1.0x+2.0, r=.98, 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=.98, SEE=3.3%).

Conclusions Despite the irregular crescentic shape of the right ventricle, this newly developed three-dimensional system and surfaceing 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.1-7 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.8 Two-dimensional echocardiographic RV volume determinations, whether based on correlations with two-dimensional dimensions,9-11 or Simpson's rule methods,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 infundibulum6,7,9-12,14 and because of difficulties in obtaining the set of two perpendicular views used in angiography9-12 and imaging them in a standardized fashion.

Three-dimensional echocardiography, which has recently been validated for determining left ventricular (LV) volume,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 vitro40,41 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.39 Although this system has been validated for the LV in vivo,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-
bated, and ventilated. A midline thoracic incision was performed, the pericardium was incised, and the heart was suspended in a pericardial cradle. To obtain actual volumes in vivo, the following model adapted from the method of Suga and Sagawa,42 as applied by Weiss et al,27 was used in which instantaneous RV volume can be measured directly using an intracavitary balloon connected to an external reservoir (Fig 1). This model isolated the RV cavity from the circulation by draining systemic and coronary sinus venous return and pumping it through an oxygenator to the left atrium and LV, which perfused the coronary arteries in a normal manner. A high-compliance latex balloon was inserted into the RV through the pulmonary artery (to fill the infundibulum and the rest of the RV), and the tricuspid valve was sewn shut. To ensure that the balloon could fill the entire RV cavity and conform maximally to its contour, the tricuspid valve chordae tendineae were cut, and the thebesian venous return was drained by a 24-gauge cannula inserted into the RV apex. The balloon was connected at the pulmonary valve level to an extracardiac vertical polyurethane column (fixed 3/4-in. internal diameter). Known amounts of fluid could be added or withdrawn through a port at the lower end of the column. Compressing a section of Tygon tubing attached to the top of the column allowed the resistance to RV contraction to be varied. During the experiment, known amounts of saline were incrementally introduced into the balloon-column system. The height of the fluid was assessed by continuous video recording of the calibrated column and subsequent off-line analysis. (The fluid was colored with blue dye for better visibility.) The RV cavity (balloon) volume was determined as the total volume in the system (balloon and column) minus the volume in the column, measured from the videotaped images. Systolic cavity volume was determined from the peak fluid level in the column, and diastolic volume was determined from the lowest level, averaged over five consecutive beats. To maintain a constant heart rate, the sinusoidal node was crushed, and the RV was paced at 80 to 90 beats per minute. Conformity of the balloon to the shape of the RV walls was confirmed by echocardiographic imaging, which displayed an interface similar to that in the native setting, with no separate space to indicate an independent balloon shape within the cavity (Fig 2). The column was sutured in at the pulmonary valve level, so that the infundibulum and RV outflow tract were included in the scanned areas and the volume measured by the balloon.

**Experimental Protocol**

Each animal was studied in a series of hemodynamic stages created by volume loading the RV. Increments of 10 to 15 mL of fluid were added at each stage until limited by RV compliance (lack of further important balloon distension) or failure.
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.\(^{39}\) 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 \texttt{RSL} 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.\(^{40}\) To determine whether linkage of results in different animals affected results, multiple linear regression analysis of three-dimensional versus actual volumes was performed (\texttt{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).
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 0.97 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±2.6 mL (P<.05; Fig 4C, bottom). Correlation was similarly good for RV ejection fraction (y=0.96x−0.03, r=.98, SEE=3.3%) (Fig 4D, top, and Table). The mean difference between three-dimensional and actual values was −4.5±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 radio- 

nuclide data because of several potential factors: (1) limitations of simplified geometric formulas in describing the complex RV, which is crescentic and asymmet- 

rical, 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 standard- 

ized imaging planes. Its accuracy for assessing the more 

symmetrical left ventricle has been previously demon- 

strated both in vivo and in vitro\textsuperscript{23-35,39}; 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 pre- 

cent study and a previous study,\textsuperscript{40} provides several other advantages.\textsuperscript{59} 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 advan- 

tages. The location data are generated in real time and 

recorded simultaneously and automatically with the 

two-dimensional images, eliminating the need for man- 

ual 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 slices40-45 (although that was not the central feature of the previous method or the major contributor to its accuracy40).

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

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**Observer variability** = 1.86 cc (4.0%)  
**Volume**

![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.](image)
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 para-apical long-axis views can be reconstructed; (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, 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 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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