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Calculation of Aortic Regurgitant Volume by a New Digital Doppler Color Flow Mapping Method: An Animal Study With Quantified Chronic Aortic Regurgitation

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Objectives. **The aim of the present study was to quantitate aortic regurgitant volume and regurgitant fraction in a chronic animal model with surgically created aortic regurgitation using a new semiautomated color Doppler flow calculation method.**

Background. **The conventional noninvasive methods for evaluating the severity of aortic regurgitation have not been accepted widely nor compared with truly quantitative reference standards.**

Methods. **Eight to 20 weeks after aortic regurgitation was surgically induced in six sheep, a total of 22 hemodynamic states were studied. Electromagnetic flow probes and meters provided reference flow data. Epicardial color Doppler echocardiographic studies were performed to image left ventricular outflow tract forward and aortic regurgitant blood flows. The new method digitally integrated spatial and temporal color flow velocity data for left ventricular outflow tract forward flow and ascending aortic regurgitant flow. The pulsed Doppler method using the velocity–time integral was also used to obtain regurgitant volumes and regurgitant fractions.**

Results. **Regurgitant volumes and regurgitant fractions by the new method agreed well with those obtained electromagnetically, whereas the pulsed Doppler method overestimated these reference data (mean [± SD] difference 0.23 ± 2.9 ml vs. 11 ± 5.8 ml, $p < 0.0001$ for regurgitant volume; mean difference 1.2 ± 7.6% vs. 19 ± 13%, $p < 0.0001$ for regurgitant fraction).**

Conclusions. **This animal study, using strictly quantified aortic regurgitant volumes, demonstrated that the digital color Doppler method provides accurate aortic regurgitant volumes and regurgitant fractions without cumbersome measurements.**

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A number of echocardiographic methods have been developed to evaluate the severity of aortic regurgitation (1–17). Success in estimating aortic regurgitant flow volumes and regurgitant fractions has been reported using pulsed Doppler methods that multiply the cross-sectional area of the great arteries or mitral annulus, or both, by the velocity–time integral at the center of the flows (5,6,10,13). However, such noninvasive methods for estimating the severity of aortic regurgitation have most often been compared with angiographic grading, which in itself depends on many technical factors and may differ substantially from those determined by quantitative flow measurements (18,19). More importantly, all of the pulsed Doppler methods assume a *flat* velocity profile during the entire duration of flow and a constant flow area. However, for low velocity flows the spatial velocity distribution is not necessarily flat (20) and the flow area is temporally changing. Therefore, the assumptions of the pulsed Doppler method (i.e., flat velocity profile and constant flow area) do not hold true. Recently, a computer-assisted semiautomatic digital color Doppler method that avoids having to rely on these assumptions has been described for calculating stroke volume and cardiac output (21–23). Because this new method digitally integrates temporal and spatial velocity assignments across the entire diameter of the ascending aorta, it should provide more reliable regurgitant flow volumes than the pulsed Doppler flow method.

The aim of the present study was to evaluate the new Doppler flow calculation method versus the pulsed Doppler method for determining aortic regurgitant volume and regurgitant fraction using a chronic animal model with strictly quantified aortic regurgitation.

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Methods

Six juvenile sheep weighing 32 to 64 kg (mean 44 ± 14) were studied. Eight to 20 weeks (mean 14) before the hemodynamic and ultrasonic studies that constitute the experimental setting for the present study, the animals had undergone open heart surgery. At that time the free edge of right coronary cusp (n = 3) or the noncoronary cusp (n = 3) of the aortic valve was incised with a radial incision under direct vision. Subsequent aortic dilation or leaflet retraction, or both, resulted in anatomic leaflet defects and failure of leaflet coaptation. All operative and animal management procedures were approved by the Animal Care and Use Committee of the National Heart, Lung and Blood Institute. Preoperative, intraoperative and postoperative animal management and husbandry methods are described in detail elsewhere (24,25). Postoperatively, the animals were maintained on digoxin and furosemide.

Electromagnetic flow probe and meter method. An electromagnetic flow probe (model EP455, Carolina Medical Electronics, Inc.) was placed around the pulmonary artery just above the pulmonary valve sinuses. Another electromagnetic flow probe (model EP455, Carolina Medical Electronics) was placed snugly around the skeletonized ascending aorta distal to the coronary ostia and proximal to the brachiocephalic trunk. Both flow probes were connected to flow meters (model FM501, Carolina Medical Electronics), and these were connected to physiologic recorders (ES 2000, Gould, Inc.) used for hemodynamic pressure recordings. Aortic and left ventricular pressures were obtained from intracavity, manometer-tipped catheters (model SPC-350, Millar Instruments, Inc.) positioned transmurally. All hemodynamic data were recorded at paper speeds of 250 mm/s. Four consecutive cardiac cycles were analyzed for each hemodynamic determination.

Calibration factors for the flow probes were corrected for the animals' hematocrit levels before each hemodynamic state, according to the manufacturer's specification. The integrals of instantaneous flows over time were determined by planimetry of the flow signal recordings. The problem of the zero baseline drift was managed as previously described (16,17), such that the baseline value for the aortic flow recording was adjusted until the forward minus the backward aortic flow volumes equaled the pulmonary forward flow volume. Coronary artery blood flow during ventricular diastole was measured in three sheep in a preliminary study. The coronary flow rate was low (0.13 to 0.23 liter/min). These values were similar to those reported by other investigators studying aortic regurgitation, and thus were considered to be negligible compared with the regurgitant volumes delineated in this study (10). Regurgitant fraction was calculated as backward aortic flow volume per minute divided by forward aortic flow volume per minute.

After baseline measurements, varying degrees of severity of aortic regurgitation were produced by altering preload or afterload, or both, using blood transfusion or angiotensin infusion, or both. The calibrations of the flow probes were readjusted before each individual hemodynamic steady state, compensating for any change in hematocrit produced by insensible fluid loss, blood loss or alteration of preload by blood transfusion, or a combination of these factors. Insensible fluid loss and associated electrolyte disturbances exacerbated by the open thoracotomy were monitored by frequent (before each individual hemodynamic study) determinations of serum electrolyte and hematocrit; aberrations were avoided by continuous infusion of lactated Ringer's solution and 5% dextrose in water supplemented with potassium and calcium, as necessary. A total of 22 hemodynamic states (two to four per animal) was obtained.

Acute animal model study. To investigate instrument and technical factors such as color gain setting, color Doppler filter setting and off-axis imaging, which may affect the flow measurements using the new color Doppler method, we surgically created aortic regurgitation in two other sheep and acutely instrumented them with the electromagnetic flow probes on both the aorta and pulmonary artery, as in the chronic animal model. We used these experiments as an opportunity to obtain orthogonal plane comparisons for calculation of forward and retrograde flows.

Color Doppler echocardiography. A Toshiba Power Vision SSA-380A was used to image aortic forward and regurgitant blood flow with a 3.75-MHz sector probe. The ultrasound probe was placed directly on the heart near the apex, and ascending aorta and left ventricular outflow tract images were obtained using apical long-axis views (Fig. 1). Color gain was adjusted to eliminate random color in areas without flow. The color Doppler wall filter of 0.05 to 0.10 m/s was used so as not to lose low velocity information. To optimize instrument settings, two aliasing velocities—0.32 and 0.72 m/s—were selected for the initial imaging of the retrograde flow in the ascending aorta, and an aliasing velocity of 0.72 m/s was used for forward flow through the left ventricular outflow tract. A narrow color sector was chosen to allow frame rates as high as 32/s. To avoid aliasing, on-line baseline shifting was performed to obtain velocities as high as 1.44 m/s for forward flows through the left ventricular outflow tract.

In the acute animal model, instrument and technical factors were investigated using the same echocardiographic system, as follows: 1) The color Doppler gain setting was changed by −2 dB from the optimal settings. 2) The color Doppler wall filter was changed from the original 5.6 cm/s to 7.8 cm/s. These alterations were chosen as being within a range wherein color Doppler images would still appear "good" in quality. 3) The position of the echocardiographic probe was taken from the apex and moved toward the base so as to obtain off-axis imaging (45° to the blood flow direction) of the aortic flow, which may be the only view obtainable clinically. 4) The orthogonal plane (five-chamber view) to the original apical long-axis view also was used to measure the forward and retrograde flows. These data were compared with flow measurements obtained by using the original settings and the electromagnetic flow meters.

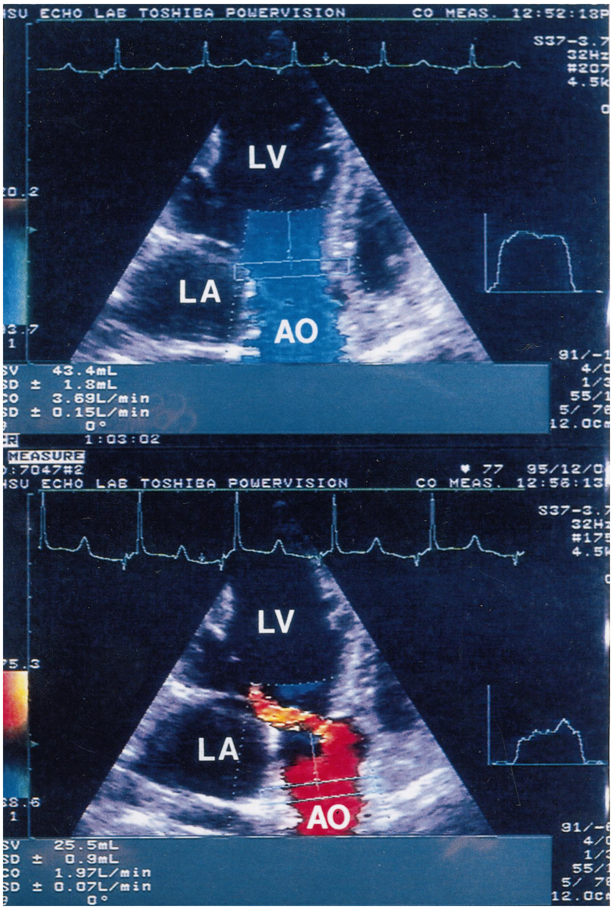


Figure 1. Examples of selected two-dimensional color Doppler images for measurements of forward flow volume in the left ventricular outflow tract (**top**) and aortic regurgitant flow volume (**bottom**). Note a flatter velocity profile (x axis = distance, y axis = velocity) shown at the right side of the top panel than that of the bottom panel. In this example, a regurgitant volume of 43 ml was automatically shown on the monitor, and a regurgitant fraction of 59% was calculated. AO = aorta; LA = left atrium; LV = left ventricle.

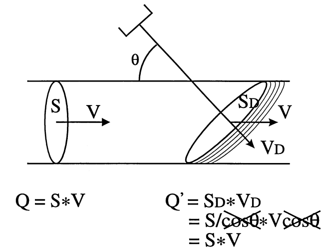


Figure 2. Schema of the principle for calculating actual flow rate (Q = V × S) using Doppler-imaged flow velocity. Note that obliquity decreases velocities by a cosine function but increases areas by the same function. Five parallel velocity interrogations in a region of interest were averaged in the system (see text). S = area; SD = Doppler area; V = velocity; VD = Doppler velocity.

Semiautomated calculation of aortic regurgitant volume and regurgitant fraction. Doppler color flow mapping images were recorded as digital cine loops (maximum 255 frames) (Fig. 1). Aortic regurgitant volumes and forward flow volumes were calculated using the automatic calculation package in Power Vision, as follows (21).

For calculating the forward flow volumes, the region of interest was placed in the left ventricular outflow tract and a total of 5 to 10 frames during each systole was selected manually from the digitally recorded cine loop (Fig. 1). For calculating aortic regurgitant volumes, the region of interest was positioned in the ascending aorta 1.5 to 2.0 cm above the aortic valve and a total of 7 to 14 frames during each diastole was selected manually from the cine loop (Fig. 1).

The size of the region of interest was 33 (width) × 5 mm (depth). Each region of interest included five discrete velocity profiles sequential in depth (thickness 1 mm) and parallel to each other across the diameter. Each discrete velocity profile included 115 Doppler velocity data points across the 33 mm of its width, which were derived from the original 42.6 scan lines at the first profile closest to the transducer and 38.1 scan lines at the fifth furthest profile using computed interpolation.

The flow rate from each one of the five velocity profiles from each still frame was calculated by integrating the Doppler interrogated velocity across the diameter in digital cine memory, assuming a half-circular symmetric flow distribution for each radius. Figure 2 shows the principle of the flow measurement using the new Doppler system; the actual flow rate (Q) is equal to the product of the velocity of the flow (V), which is parallel to the direction of the vessel, times the area (S), which is the area of flow perpendicular to the vessel. This flow rate is also equal to the product of color Doppler-determined velocity ($V_{Doppler} = V \times \cos\theta$) and the corresponding area of flow ($S_{Doppler} = S/\cos\theta$) perpendicular to the Doppler-determined velocity; the flow rate calculated by using Doppler-determined velocities is $V\cos\theta \times S/\cos\theta = V \times S = Q$. Note from Figure 2 that any obliquity to the flow direction increases flow cross-sectional area in proportion to the decrease in computed velocity.

All color Doppler acquisitions for the chronic experiments were obtained parallel to flow or at angles <10°. Imaging was adjusted to include a long segment of the left ventricular outflow tract and the aorta to avoid obliquity in elevation. However, when we deliberately performed off-axis imaging in the acute experiments, we used the angle correction cursor that corrected the calculated flow volumes for velocity decrease and flow area increase caused by the Doppler angle. Because this ultrasound imaging system assumes a circular flow area (actually a half circle for each radius), the imaged radius is *squared* automatically, and thus the flow area is overcorrected by a factor of ($1/\cos\theta$) in the long-axis view of the aorta. The angle correction software equipped with the prototype ultrasound system was modified for multiplying by $\cos\theta$, to avoid overestimation of the actual flow volumes when imaging the long-axis view obliquely.

Five different flow rates from the five sequential velocity profiles in the depth domain within the region of interest of each frame were then averaged to determine the representative flow rate for that still frame. The same calculation of the flow rate was performed consecutively for each frame during the systolic or diastolic time of flow. All of these flow rates during the selected period of flow were added together and multiplied by the time interval duration to obtain the flow volume during that particular systole or diastole (21). This flow volume per beat was multiplied by heart rate to provide the averaged flow volume per minute. Results of the calculations of flow volumes per beat and minute (with standard deviations) were automatically displayed by the system, as was the velocity profile (Fig. 1).

Averages of 6.4 frames/systole and 8.6 frames/diastole were available based on a mean heart rate of 95 beats/min. Four determinations of both forward and regurgitant flow volumes at each hemodynamic state were averaged.

Pulsed Doppler method. The pulsed Doppler mode was used to obtain central modal velocities in both the ascending aorta and left ventricular outflow tract. Two-dimensional imaging from conventional parasternal views was used to measure the maximal diameters of the ascending aorta and left ventricular outflow tract. We measured the velocity–time integrals of the pulsed Doppler velocity by tracing the modal velocity with a track ball system (26). The flow area was calculated as ($\pi \times$ Diameter^2/4) of the ascending aorta and left ventricular outflow tract, assuming circular geometry as previously reported. Then the regurgitant flow volumes and forward flow volumes were calculated as the product of the velocity–time integral and the flow area (5,6,13,26–32). Measurements were performed during four cardiac cycles at each hemodynamic state, and the data were averaged. Regurgitant fraction was calculated as the ratio of the regurgitant volume to the forward flow volume.

Both regurgitant volumes and regurgitant fractions determined by the new automated color Doppler method and the pulsed Doppler velocity–time integral method were compared with reference corresponding data obtained by the electromagnetic flow meter method.

Interobserver variability. To evaluate the effect of observational variability on the measurement of forward and retrograde flow volumes, 10 randomly selected flow conditions were analyzed with the same ultrasound system by two independent observers who had no knowledge of the images selected or the results obtained by the other observer or the flow meter data.

Statistical analysis. Data are presented as mean values $\pm$ SD. Correlations between continuous variable data were determined by linear regression analyses. Statistical significance was defined at $p < 0.05$. The relation and agreement between the electromagnetically determined regurgitant volumes and regurgitant fractions and those calculated by echocardiography were tested according to the method of Bland and Altman (33). In addition, because multiple points were used from the same animal, multiple regression analyses were used to examine the relations of data within sheep. To do this, we created the data matrix in the spread sheet of a statistical computer program (Stat View 1988, Abacus Concepts, Inc.) using dummy variables as columns to encode the different sheep and we used the multiple regression function of Stat View (34,35).

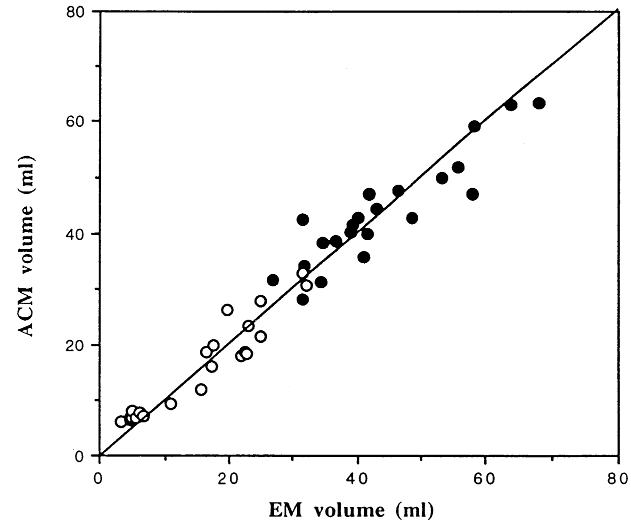


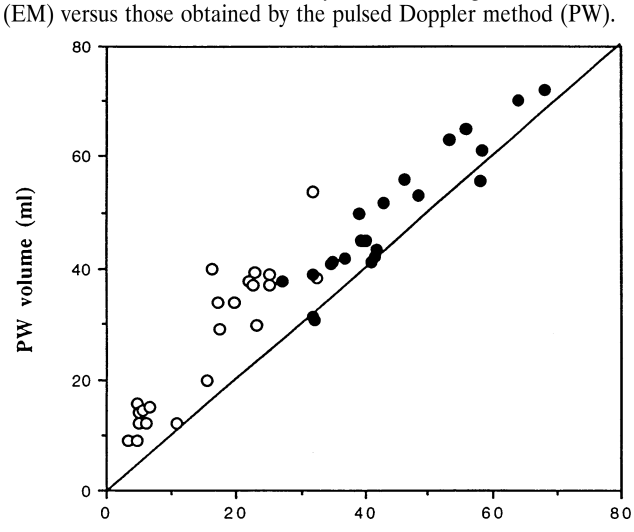
Figure 3. Linear regression analyses of forward flow volumes through the left ventricular outflow tract **(solid circles)** and aortic regurgitant volumes **(open circles)** obtained by the electromagnetic flow meters (EM) versus those obtained by the automated cardiac flow measurement (ACM) method.

Results

Severity of aortic regurgitation. Aortic regurgitant volumes and regurgitant fractions obtained by the electromagnetic flow meters were within clinically relevant ranges from 3.3 to 34 ml/beat (average 17 $\pm$ 10) and from 10% to 60% (average 34 $\pm$ 17%), respectively. Heart rates ranged from 64 to 125 beats/s (average 95 $\pm$ 23).

Estimation of regurgitant volumes and regurgitant fractions. Regurgitant volumes by the new digital method agreed well with those obtained by the electromagnetic flow meters, whereas the pulsed Doppler method overestimated these

Figure 4. Linear regression analyses of forward flow volumes through the left ventricular outflow tract **(solid circles)** and aortic regurgitant volumes **(open circles)** obtained by the electromagnetic flow meters (EM) versus those obtained by the pulsed Doppler method (PW).



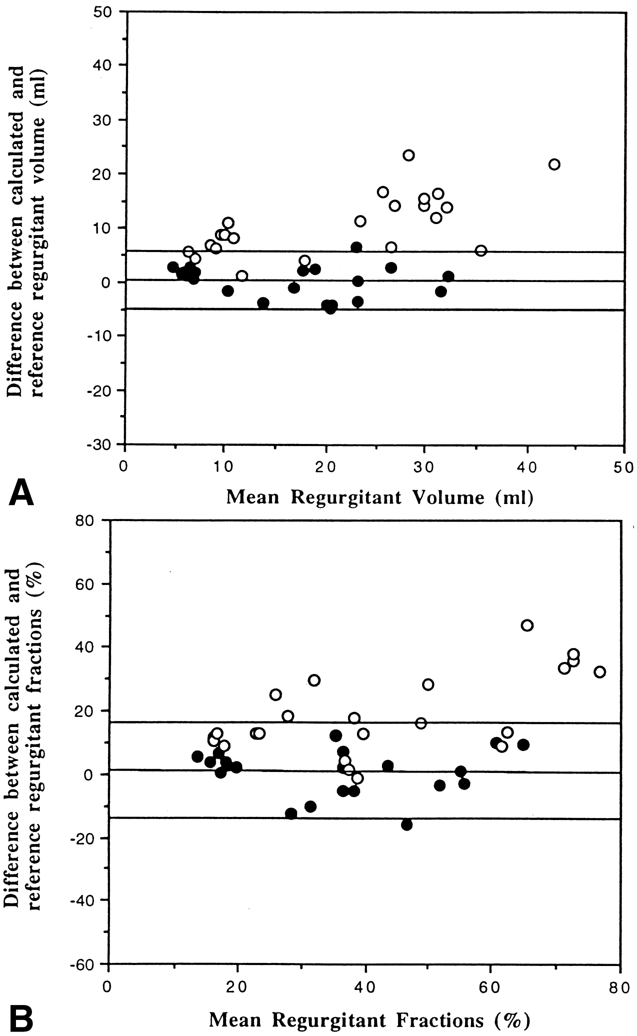


Figure 5. Agreement between aortic regurgitant volumes (**A**) and regurgitant fractions (**B**) obtained by the electromagnetic flow meters compared with the automated cardiac flow measurement method (**solid circles**) and the pulsed Doppler method (**open circles**), according to the method of Bland and Altman (33). Mean value (**center line**) and mean value ±2 SD (**outer lines**) are shown only for automated cardiac flow measurement (see text).

reference data (y = 0.88x + 2.1, r = 0.95, p = 0.0001, SEE 2.8 ml, mean difference 0.23 ± 2.9 ml vs. y = 1.3x + 6.1, r = 0.93, p = 0.0001, SEE 5.2 ml, mean difference 11 ± 5.8 ml) (Fig. 3, 4 and 5A). When multiple regression analysis was used to eliminate the effects of the repeated measurement in the same animal, a weaker but still similar relation between the calculated and electromagnetically obtained regurgitant volumes was found (y = 0.85x + 2.3, r = 0.92, p = 0.0001 for the color Doppler method vs. y = 1.2x + 6.5, r = 0.89, p = 0.001 for the pulsed Doppler method).

Forward flow volumes were also better estimated by the new digital method than by the pulsed Doppler method when compared with values obtained from the electromagnetic flow meters (y = 0.77x + 9.8, r = 0.91, p = 0.0001, SEE 4.1 ml, mean difference −1.1 ± 6.2 ml vs. y = 1.2x − 2.0, r = 0.89, p = 0.0001, SEE 7.1 ml, mean difference 9.8 ± 9.8 ml) (Fig. 3 and 4). The degree (percentage) of overestimation of the regurgitant flow volumes obtained by the conventional pulsed Doppler method was more severe than that for forward flow volumes, probably due to the irregular velocity profiles of the former, as seen in Figure 1 (71% vs. 14%, p < 0.001).

As a result of these measurements, aortic regurgitant fractions by the new semiautomatic digital method agreed well with electromagnetically obtained reference regurgitant fractions, whereas the pulsed Doppler method overestimated the reference values substantially (r = 0.90, p = 0.0001, SEE 7.5%, mean difference 1.2 ± 7.6% vs. r = 0.87, p = 0.0001, SEE 12%, mean difference 19 ± 13%) (Fig. 5B).

Instrument and technical factors for data acquisition. Figure 6 shows the results from different instrument and technical settings obtained from the acute animal model of aortic regurgitation. There were no significant differences (all p > 0.2) in either forward or retrograde flow measurements obtained by the new color flow method using different color gain, wall filter, orthogonal imaging or altered angle. Probably because color gains and wall filters were selected to provide "acceptable" color image quality, these changes did not provide significant differences in the results. Both the optimal setting and all four altered settings agreed well with electromagnetic flow meter values (mean difference 0.03 ± 0.43 ml/beat for the optimal setting, −0.23 ± 1.7 ml/beat for the lower color gain, −1.4 ± 2.0 ml/beat for the higher wall filter, 0.03 ± 0.82 ml/beat for off-axis imaging and 0.08 ± 1.8 ml/beat for the orthogonal plane, all p > 0.2).

Observer variability. There was excellent agreement between the two independent observers' measurements of aortic regurgitant volumes and regurgitant fractions using the new digital color Doppler method (r = 0.94, difference 2.5 ± 3.4 ml and r = 0.91, difference 4.2 ± 3.5%, respectively).

Discussion

In the present study, using strictly quantified chronic aortic regurgitation in an animal model, the semiautomatic calculation of regurgitant volumes in the ascending aorta and regurgitant fractions was proved to be more reliable than the pulsed Doppler method. In addition, this new method did not require complicated or cumbersome off-line measurement techniques.

Aortic regurgitant volume and regurgitant fraction. Many noninvasive methods have been proposed for evaluating the severity of aortic regurgitation (1–17). Most Doppler echocardiographic methods, such as pressure half-time, continuous wave regurgitant velocity deceleration slope and pulsed Doppler flow velocity pattern in the descending aorta, provide only indirect estimation of regurgitant volume and regurgitant fraction. These have been compared with angiography, which in itself is only semiquantitative (18,19). Use of the flow convergence phenomena also provides only semiquantitation of regurgitant volumes and regurgitant fractions (11,14). To estimate aortic regurgitant volumes, flow convergence methods require complicated techniques such as multiplicative

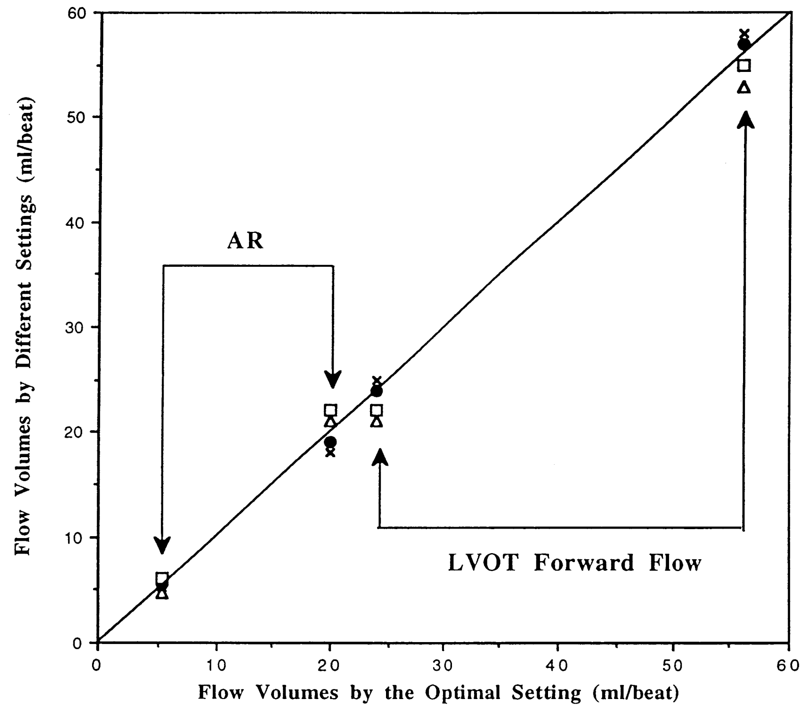


Figure 6. Linear regression analysis of the flow volume measurements for the original optimal setting versus the different settings for two animals—one with mild and one with more severe aortic regurgitation. Values using the optimal settings were compared with those using orthogonal planes (biplane) and three other altered settings or acquisition techniques (see text for details). LVOT = left ventricular outflow tract. **Solid circles** = off-axis; **crossmarks** = gain; **squares** = biplane; **triangles** = filter.

curve fits of the axial centerline velocity/distance curves and integration of the regurgitant flow rate over time (16). Using the conventional hemispheric isovelocity surface assumption of the flow convergence, a priori selection of an optimal aliasing velocity for imaging the flow convergence toward the regurgitant orifice is required to calculate the peak regurgitant flow rate (17). Then regurgitant volumes are calculated as the product of the resulting peak regurgitant flow rate and the ratio of the velocity–time integral and the peak velocity of the continuous wave Doppler-determined regurgitant orifice velocity (17). Thus, the flow convergence method requires two-step calculations to estimate regurgitant volumes. The angle between the Doppler ultrasound interrogation and the actual blood flow may cause inaccuracies with this method (17). In addition, to obtain regurgitant fraction, it is still necessary to obtain forward flow volume accurately, which is not possible using the current flow convergence methods. In contrast, the present Doppler technique provides direct estimation of both regurgitant flow volumes and forward flow volumes (and the regurgitant fraction as the ratio of the former to the latter).

Problems of pulsed Doppler methods for flow volume estimation. Alternatively, pulsed Doppler methods have been reported to be reliable for calculating forward flow volumes through the left ventricular outflow tract, pulmonary artery and mitral valve, and thus have been widely used experimentally and clinically (5,6,10,13,26–32,36). Most pulsed Doppler methods use a single (central) velocity as being representative of the entire velocity flow field, assuming a spatially *flat* velocity profile during the entire period of flow. However, the spatial distribution of the flow velocity depends on many factors, including the size and geometry of the flow area and flow volumes (20). Thus, assuming spatially flat velocity profiles during the entire period of flow is an oversimplification, especially if the flow rate is low and the flow area is relatively large (20). Recently, it has been reported that the pulsed Doppler method substantially overestimates inflow volumes through simulated mitral valves (37,38). In our study, overestimation by the pulsed Doppler method was smaller for forward flow than for regurgitant flow. This may be explained by the spatially flatter velocity distribution of the forward flow in the left ventricular outflow tract, as compared with the velocity profile of the retrograde flow in the ascending aorta (Fig. 1). The pulsed Doppler method, which we used to calculate aortic regurgitant volumes, was not the conventional method used clinically, which subtracts mitral inflow from aortic outflow (6,13). We chose the pulsed Doppler method because it could be implemented from one view, and by comparison, it clearly shows the advantage of the new digital method (i.e., the importance of using not just a single central velocity, but also all of the spatial velocity assignments across the flow diameter).

Another possible factor related to the limitation of the pulsed Doppler method observed in this study is the use of the constant (maximal) flow area during the cardiac cycle for calculating flow volumes. Considering the temporal change of the area of flow during the cardiac cycle (39), this method may cause overestimation of the retrograde flow volumes when flow is calculated from the product of the maximal area of the retrograde flow and the velocity–time integral of the central flow velocity. It has been suggested that changes in the flow area may cause substantial variability of flow volume calculations, especially for flows through mitral and pulmonary valves (26). Because of this variability, cumbersome techniques for averaging the flow area throughout the cardiac cycles are required to obtain accurate mitral flow volumes (28). Kitabatake et al. (5) reported in patients successful estimation of aortic regurgitant fractions using the conventional pulsed Doppler method for calculating left and right ventricular outflow tract flows. However, temporal variability of the flow area may induce inaccuracy, especially for the right ventricular outflow tract (26,28). In contrast, this new semiautomatic method can calculate flow volumes using different areas of flow during the cardiac cycle (at least four to six different flow areas in a cardiac cycle for both forward and retrograde flows) without a cumbersome manual procedure or calculation of flow area.

New methods for cardiac flow volume calculation. Recently, new color Doppler methods using principles similar to the present method for calculating cardiac outputs have been described (37,38,40–44). Some of these reports used digital color Doppler flow mapping data for integrating surface velocity vectors over a spheric flow surface at a fixed distance from the probe (37,40–44). This spheric approach does not require angle correction, whereas in the present method, the angle between the ultrasound beam and the flow direction differed very slightly at different points in the rectangular region of interest. However, the error caused by this angle would be at most 1% to 5% at the two lateral edges of the region of interest at a depth of 5 cm. Errors at other, noncentral locations near the center of the region of interest should be even smaller. Also in clinical settings, the depth of the region of interest would be >5 cm, and thus the total effect of this angle on the flow volume calculation would be minimal. To obtain high packet size and accurate Doppler velocity information, the previously described method required cumbersome electrocardiographic triggering because of low frame rates. Although the concept is reasonable, the complicated averaging procedure may hinder its clinical applicability. In contrast, in the present method, parallel processing was used to obtain high frame rates and high packet sizes to improve temporal resolution and velocity accuracy simultaneously, permitting semiautomatic calculation of flow volumes. Once a digital cine loop was recorded, the regurgitant volume as well as cardiac output was easily determined by the software in the echocardiographic system in <20 s.

Magnetic resonance imaging has also been reported to be successful in obtaining actual velocity information and aortic regurgitant volumes (39). Flow measurement using magnetic resonance imaging does not have the angle dependency of Doppler interrogation, and direct velocity profiles can be used for flow calculations. With this method, there is no restriction in selecting windows for obtaining real velocity information in the ascending aorta. However, current magnetic resonance imaging methods do have the limitations of low frame rates and long acquisition times as well as high costs and the requirement of special facilities.

Study limitations. In addition to the inherent limitations of experimental studies which we have discussed previously (16,17), the present method assumed axisymmetric flow (i.e., velocity information from one diameter was assumed to be representative of the entire flow area velocity). This assumption may be an oversimplification, especially if the regurgitant orifice is not located in the center of the aorta and the measured radius is close to the regurgitant orifice. However, the distance between the regurgitant orifice and the area of the ascending aorta that we measured (1.5 to 2.0 cm above the aortic valve) should be far enough away from the orifice to assume an axisymmetric velocity distribution of the retrograde flow. During our initial experience, we rotated the probe to determine the proper level for obtaining a spatially uniform velocity distribution and found this to be the case at any interrogation angle at a distance of 1.5 to 2.0 cm above the aortic valve. In the present study, orthogonal imaging (five-chamber view) of the aorta provided similar retrograde flow volumes to those obtained using the original apical long-axis view. Thus, our results should be applicable for aortic regurgitant volume estimation in the clinical setting, unless the aorta is severely tortuous or distorted. In the left ventricular outflow tract, skewed forward velocity profiles with higher velocities along the anteroseptal than along the posterolateral portion and proximal aorta have been reported (41–43). However, using two orthogonal views for imaging the left ventricular outflow tract provided similar results, probably because of averaging over the entire axial flow velocity fields in our animal study and in a clinical one by Sun et al. (22).

Other important limitations are instrument related and technical factors, which may affect the flow measurements by any color Doppler method. Recently, in vitro and clinical studies investigating instrument factors showed the effect of the color gain setting, wall filter, transmission frequency and depth of the region of interest on the flow volume measurement using this new method (22,45,46). It may not always be possible to use such optimal settings clinically as we used in this animal model, because of deeper areas of interest or poor image quality, or both. Also, for mild aortic regurgitation, the use of higher wall filters may cause underestimation of the actual regurgitant volumes due to loss of lower velocities near the aortic wall (45). When apical views are not available or suboptimal for obtaining clear images of retrograde flow during diastole, other echocardiographic windows, including right parasternal or transesophageal multiplane views, may be required for color Doppler measurement of aortic retrograde flows. For such views, angle correction is required if there is a significant angle between the Doppler interrogation and the direction of the retrograde blood flow. However, based on the results from the present study, the angle correction software for the color Doppler method we used does provide reliable estimation of the actual off-axis forward and retrograde flow volumes, suggesting the potential for clinical applicability of this new method. Also, because it was not possible to change the size of the region of interest at the time of our study, one might expect contamination of other flows within the region of interest. In this study, however, we were able to obtain a suitable area of flow for the region of interest by selecting appropriate depths, thus avoiding contamination from flow in adjacent areas. Also, a software release in 1997 makes it possible to adjust the size of the region of interest, obviating the potential problem.

Conclusions. This animal study, using strictly quantified regurgitant flow data, demonstrates that the newly developed digital cardiac flow calculation method provides more accurate quantitative information about aortic regurgitant volumes and regurgitant fractions than the pulsed Doppler methods.

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