Quantification of Mitral Regurgitation by Integrated Doppler Backscatter Power

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Objectives. We attempted to determine whether continuous wave Doppler backscatter power could be used to quantify mitral regurgitation.

Background. The power of a Doppler backscatter signal is proportional to the number of scatterers insonated and, hence, to the moving volume of blood. The relative power of the continuous wave Doppler signals from mitral inflow and aortic outflow is therefore proportional to the relative volumes of blood in motion.

Methods. Computer postprocessing was used to derive the relative power of the Doppler backscatter signal from the intensity of the pixels within the spectral display of anterograde aortic and mitral flow. The power ratio was used to calculate the regurgitant fraction in 20 patients (mean age 61.4 years) with mitral regurgitation. This Doppler regurgitant fraction was compared with

Although Doppler echocardiography provides a sensitive and specific means of detecting mitral regurgitation (1), the quantification of mitral regurgitation remains problematic (2,3). Attempts to quantify regurgitant volumes from Doppler color flow maps of the regurgitant jet have proved disappointing (4). This is partially because of technical limitations in image quality, but more fundamentally because of the variable relation between regurgitant volume and the regurgitant jet area (5). Quantification of regurgitation from the combination of Doppler measures of transvalvular velocities and two-dimensional valve areas (6-11) has proved difficult (12,13), mainly because of the noncircular shape of the mitral valve and diastolic variation in its area (14). Such measurements are also time-consuming and cumbersome. Continuous wave Doppler ultrasound is used to detect velocity and direction of blood flow, but no use is made of the signal power, which contains

that derived from angiographic left ventricular volume and thermodilution cardiac output. In addition, 12 normal control subjects were studied by the Doppler method.

Results. Mean (±SD) catheterization regurgitant fraction was 0.50 ± 0.26, and mean Doppler regurgitant fraction was 0.47 ± 0.25 (r = 0.89). The limits of agreement between the two methods by Bland-Altman analysis were −0.21 to +0.27. In normal control subjects with an expected regurgitant fraction of close to zero, mean Doppler regurgitant fraction was 0.03 ± 0.05.

Conclusions. Doppler backscatter power from mitral and aortic inflow provides a new and accurate method for quantifying mitral regurgitation.

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proportional to the velocity of motion. If the blood is moving at a velocity \( v \) and an angle \( \theta \), relative to the ultrasound transducer, the frequency of the scattered ultrasound is Doppler shifted by an amount

\[
f_D = \frac{2f_0 v \cos \theta}{c},
\]

where \( f_D \) is the Doppler shift, \( f_0 \) the ultrasound frequency and \( c \) the speed of sound (6). The power of this Doppler signal is proportional to the volume of moving blood insonated. If a volume of blood is moving at a range of velocities, each giving rise to a particular Doppler frequency, the sum of the powers of the individual Doppler frequencies will be proportional to the volume of blood in motion (16):

\[
V \propto \sum \frac{f_{D,n}}{f_{D,n}} P_{n}
\]

where \( P_n \) is the power of each \( n \)th Doppler frequency component produced by the moving volume.

If the hematocrit, the degree of red blood cell aggregation and the degree of turbulence are similar in two regions of flow, then the relative power of the continuous wave Doppler backscatter signals from these regions of flow will be proportional to the relative moving volumes insonated. If it is assumed that the volume of blood in motion has a constant relation to the volume of flow, then integrating the Doppler backscatter power over the duration of flow will provide a result proportional to the volume of blood flowing across the valve.

Mitral inflow and aortic outflow are both regions of laminar flow, with similar hemodynamic conditions, therefore it is reasonable to assume that the backscatter coefficients from both regions will be similar. If both flows are insonated from the cardiac apex, then the attenuation of both signals will be identical because the valves are equidistant from the cardiac apex. Hence the power of the Doppler backscatter signals will be proportional to the volume of mitral inflow and aortic outflow, respectively, and can be used to calculate the regurgitant fraction.

### Patients

Two groups were studied. There were no patients with echocardiographic evidence of aortic or mitral stenosis. Group 1 included 21 patients with mitral regurgitation. All patients in this group had a Doppler echocardiogram performed immediately before cardiac catheterization. One patient with severe mitral regurgitation in this group was excluded because left ventricular volumes could not be obtained from the left ventriculogram owing to an indistinct ventricular silhouette. The details and hemodynamic variables for the remaining 20 patients in group 1 are presented in Table 1. Group 2 included 12 control subjects who did not undergo cardiac catheterization, had a normal heart on physical examination, no abnormality on two-dimensional or M-mode echocardiography and no mitral or aortic regurgitation on Doppler echocardiography.

### Table 1. Patient Characteristics, Steady State Information and Regurgitant Fractions

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<th>Gender</th>
<th>Etiology of MR</th>
<th>Rhythm</th>
<th>HR (beats/min)</th>
<th>SBP (mm Hg)</th>
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<td>±15.9</td>
<td>±18.5</td>
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p value 0.69 0.08 0.76 0.25

A Fib = atrial fibrillation; Cath = at time of cardiac catheterization; DBP = diastolic blood pressure; Echo = at time of echocardiogram; F = female; HR = heart rate; M = male; MVP = mitral valve prolapse; Prosthetic = prosthetic mitral valve; Pt = patient; RF = regurgitant fraction; SBP = systolic blood pressure; SR = sinus rhythm.
Echocardiographic study. The mitral and aortic valves were insonated by continuous wave Doppler from the cardiac apex with the patient in the left lateral position (Acuson 128 ultrasound machine with a 2.0-MHz continuous wave transducer). The lowest filter setting (500 Hz, corresponding to a velocity ~16 cm/s) was used, so that signals from cardiac wall motion were eliminated while preserving signals from low velocity flow as much as possible. The Doppler postprocessing curve was set such that there was a linear relation between the 35-dB dynamic range and the gray-scale intensities of the spectral display. The ultrasound machine used did not have an autogain function on the spectral display and did not fill in dropout to enhance the display's appearance. The mitral inflow was insonated first, and the Doppler was gain set to optimize the spectral display. Once set, the Doppler gain and velocity scale were not readjusted during the recording of aortic outflow. The spectral display of the Doppler signal recorded from both valves represented the maximal velocity of flow and signal intensity obtainable during careful and persistent adjustment of beam direction (Fig. 1). Accurate recording of Doppler backscatter power requires the same meticulous adjustment of beam direction as that used to record the maximal systolic pressure gradient across a tricuspid or stenotic aortic valve. Both mitral and aortic flow were recorded on VHS videotape, and the images were processed off-line.

Off-line image processing. The Doppler signal was processed using a Hewlett Packard Vectra 386 personal computer containing a PCVISION Plus frame grabber (Imaging Technologies Inc.). The stored images consist of 512 X 480 arrays of 8-bit pixels. The images were acquired and processed using JAVA Version 1.40 video analysis software (Jandel Scientific). The Doppler backscatter power was derived from the pixel intensities of the spectral display of the Doppler signal. False color was added to the gray-scale display to enable subtle changes in gray-scale intensity to be detected. This process did not alter the intensities of the original signal. For patients in sinus rhythm the tape was reviewed, and five cycles were selected that showed maximal gray-scale intensity for diastolic mitral flow and systolic aortic flow, respectively. Postectopic beats were excluded. In patients with atrial fibrillation, 10 consecutive cycles of mitral and aortic flow were averaged. The envelope of anterograde flow was outlined by hand, excluding valve opening and closure signals. Next, the intensity of all pixels within this area of interest was calculated. Correction
was then made for the logarithmic compression of the data. Because of the linear relation between the compressed signal power and gray-scale intensity, the sum of the corrected pixel intensities was directly proportional to the integrated power of the Doppler backscatter signal. From the ratio of signal power of aortic and mitral flow the regurgitant fraction (RF) was calculated:

$$RF = \frac{\text{Mitral flow} - \text{Aortic flow}}{\text{Mitral flow}} = 1 - \frac{\text{Aortic flow}}{\text{Mitral flow}}$$

$$= 1 - \frac{\text{Summed aortic corrected pixel intensities}}{\text{Summed mitral corrected pixel intensities}}$$

Invasive study. Right and left heart catheterization was performed immediately before angiography. Forward flow was measured by thermodilution. Five measurements were taken, averaged and then divided by the average heart rate to calculate forward stroke volume. Stroke volume was measured from a single-plane left ventriculogram using the area-length method (20). No patient had a regional wall motion abnormality. Ectopic and postectopic beats were excluded.

Statistics. The variables measured were continuous, with the central tendency expressed as arithmetic mean and dispersion as standard deviation. The normality of data distribution was confirmed using the Shapiro-Wilk statistic. Differences in mean values were evaluated for statistical significance using a two-tailed Student t test. The Pearson coefficient was used to correlate cardiac catheterization and Doppler measurements of regurgitant fraction. Agreement between the Doppler and cardiac catheterization methods of assessing regurgitant fraction was assessed by the Bland-Altman method (21).

Results

From 20 patients with mitral regurgitation the average regurgitant fraction by cardiac catheterization was 0.50 ± 0.26, and the mean Doppler regurgitant fraction was 0.47 ± 0.25 (Table 1). Correlation between invasive regurgitant fraction and Doppler regurgitant fraction was $r = 0.89$ (Fig. 2). Agreement between the catheterization and Doppler estimates of regurgitant fraction was assessed by the Bland-Altman method (21); the Doppler estimate exceeded the catheterization estimate by 0.032 on average (bias). The standard deviation of the difference between the two estimates was 0.25; hence the limits of agreement (bias ± 2 SD) were -0.21 to +0.27 (Fig. 3). In 12 normal control subjects with an expected regurgitant fraction of zero, mean Doppler regurgitant fraction was 0.03 ± 0.05 (Fig. 4).

Discussion

Present study. Our method provides an accurate, noninvasive evaluation of the severity of mitral regurgitation. The measurement of Doppler backscatter power overcomes the technical and practical limitations of other echocardiographic methods. We found it especially useful in patients with poor image quality, in particular in those with prosthetic valves in whom the assessment of severity in terms of the size of the regurgitant jet may be misleading, and in patients with acute severe mitral regurgitation in whom the jet area recorded by color flow mapping may be smaller than expected. As such we believe that it is a valuable addition to the noninvasive evaluation of mitral regurgitation.

Because no commercially available ultrasound machine directly displays Doppler backscatter power, we devised a

*The power of the raw Doppler signal was compressed to the 35-dB range, then converted linearly to a 256-level gray-scale display. Therefore, relative to the lowest intensity gray-scale level, each nth intensity level represents a relative signal power of (1.032)$^{n-1}$. 
method to derive the relative Doppler backscatter power from the pixel intensities of the spectral display. This method was possible because the ultrasound machine used in this study generated a linear display of the logarithmically compressed Doppler signal power without enhancing the spectral display either by filling, in the case of dropout, or by use of an autogain function. Our methodology would have been more precise if a direct digital output of the raw Doppler signal had been available for analysis. If a manufacturer were to arrange the direct display of Doppler backscatter power, a simple on-line calculation of regurgitant fraction could be made.

We found good correlation ($r = 0.89$) with the reference standard provided by angiographic and thermodilution data. The concept of a reference standard does, however, have the effect of attributing all error and lack of correlation to the method under examination. Angiography and thermodilution are known to be prone to substantial errors (22), which must have made a significant contribution to the reduction of the correlation between this reference standard and the Doppler method tested. This problem is greater in the presence of atrial fibrillation because of the limited number of beats that can be measured for estimation of angiographic volume and the possibility of hemodynamic variation causing variation in the regurgitant volume. We attempted to minimize these difficulties by performing our Doppler studies immediately before cardiac catheterization. The Bland-Altman analysis displays the difference between the Doppler and catheterization assessments of regurgitant fraction. However, in view of the limitations of the accuracy of the catheterization assessment mentioned earlier, it is uncertain how much of this difference can be attributed to the errors of each method. Further testing of the Doppler power method in the clinical environment is therefore required before accepting it as an established technique for the assessment of the severity of mitral regurgitation.

Theoretic implications. Whether the scattering of ultrasound by blood is conceptualized as scattering by particles (15,23), as a fluctuating continuum (24) or as a combination of both (19), the scattered ultrasonic power is a function of the intensity of the transmitted ultrasound, the backscatter coefficient of the blood and the volume of the blood insonated (15,23-25). Thus, the relative power from two regions of flow with similar hemodynamic conditions will be determined by the relative volume of moving blood in the two regions, assuming no difference in ultrasonic attenuation. Although a full theory of anterograde flow through a valve orifice has not yet been worked out, it is reasonable to apply some of the concepts used to analyze the behavior of jets through stenosed and incompetent valves to normal anterograde flow (5). In particular, the blood flowing through an orifice has a momentum, defined by the volume of blood passing through it at any instant, and its velocity (26). This momentum is dissipated as the moving blood mixes with blood already present in the receiving chamber and slows down. The size of this moving column of blood is determined by its momentum. This relation is complicated by a number of factors, including the chamber’s walls. Nevertheless, our results indicate that the volume of the moving columns of blood, detected by the ultrasound beam, is related to both the orifice area and the velocity of flow through the orifice. This relation between the number of moving scatterers detected by the beam and the volume of anterograde flow across the valve requires further investigation.

Other studies. Other workers (27) have used amplitude-weighted mean velocities to quantify mitral regurgitation. This method is partially based on the false assumption that the backscatter amplitude and not the power is proportional to the blood volume insonated. It is true that for a fixed volume of blood the scattering power is not proportional to the number of red blood cells or hematocrit because of variations in the packing of red blood cells with a changing hematocrit. However, if the hematocrit remains constant and the blood volume is altered, then the power of the backscatter signal will change in proportion to the volume, if other factors are held constant (25). In addition, the multiplication of velocity and amplitude does not seem theoretically justified nor is this product dimensionally meaningful.

It is common practice for those interpreting continuous wave Doppler recordings to use the signal intensity of the regurgitant flow as an empiric measure of the severity of regurgitation. Indeed a semiquantitative method for assessing mitral regurgitation from the intensity of the regurgitant jet has been developed (28). However, we chose not to use the signal intensity of this turbulent regurgitant flow because turbulent flow increases the backscatter coefficient of blood, producing a high power signal (6,18,23) not directly comparable with a signal from a region of laminar flow. We excluded patients with mitral stenosis for the same reason: It would not be valid to compare turbulent mitral inflow with laminar aortic outflow. However, estimates of regurgitant fraction in patients with prosthetic valves in our study did not vary systematically from those with native valves. We therefore concluded that
localized turbulence in an unobstructed valve is not sufficient to invalidate our method of measurement.

Summary. We used the power of the continuous wave Doppler backscatter signals from aortic and mitral flow to accurately estimate regurgitant fraction in isolated mitral regurgitation. Further clinical testing of this method is now required to determine its robustness and to further elucidate some aspects of its theoretic basis.

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References