Practical Value of Cardiac Magnetic Resonance Imaging for Clinical Quantification of Aortic Valve Stenosis
Comparison With Echocardiography

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Background—Valvular pathology can be analyzed quickly and accurately through the use of Doppler ultrasound. For aortic stenosis, the continuity equation approach with Doppler velocity-time integral (VTI) data is by far the most commonly used clinical method of quantification. In view of the emerging popularity of cardiac magnetic resonance (CMR) as a routine clinical imaging tool, the purposes of this study were to define the reliability of velocity-encoded CMR as a routine method for quantifying stenotic aortic valve area, to compare this method with the accepted standard, and to evaluate its reproducibility.

Methods and Results—Patients (n=24) with aortic stenosis (ranging from 0.5 to 1.8 cm²) were imaged with CMR and echocardiography. Velocity-encoded CMR was used to obtain velocity information in the aorta and left ventricular outflow tract. From this flow data, pressure gradients were estimated by means of the modified Bernoulli equation, and VTIs were calculated to estimate aortic valve orifice dimensions by means of the continuity equation. The correlation coefficients between modalities for pressure gradients were \( r = 0.83 \) for peak and \( r = 0.87 \) for mean. The measurements of VTI correlated well, leading to an overall strong correlation between modalities for the estimation of valve dimension (\( r = 0.83 \), by means of the identified best approach). For 5 patients, the CMR examination was repeated using the best approach. The repeat calculations of valve size correlated well (\( r = 0.94 \)).

Conclusions—Velocity-encoded CMR can be used as a reliable, user-friendly tool to evaluate stenotic aortic valves. The measurements of pressure gradients, VTIs, and the valve dimension correlate well with the accepted standard of Doppler ultrasound. (Circulation. 2003;108:2236-2243.)

Key Words: magnetic resonance imaging • aorta • stenosis • echocardiography • valves
MRI techniques that can be performed rapidly and reproducibly by experienced CMR technologists for measuring aortic valve dimensions. The point of this study is to demonstrate that one can calculate values for aortic stenosis by CMR that are analogous to those obtained by echocardiography. For a series of patients with varying severities of aortic valve stenosis, the valve orifice area was analyzed by using VTIs in the continuity equation approach analogous to that used in echocardiography. The CMR data were compared directly with data acquired and analyzed independently by an experienced sonographer under routine imaging conditions. Characterized by methodological accuracy and reproducibility, the data illustrate excellent concordance between aortic valve pressure gradients, VTIs, and aortic valve orifice areas determined by echocardiography and CMR.

Methods

Patient Population

Patient volunteers were selected at random by an experienced sonographer and nurse coordinator, based on retrospective review of reports of clinically indicated echocardiograms. Patients were required to have documented aortic valve disease (mild, moderate, or severe stenosis) by echocardiography and to be clinically stable without symptoms that would have precluded CMR. The only exclusion criteria were those for general CMR suitability and subvalvular outflow tract obstruction. The protocol was approved by the local institutional review board, and all patients gave informed consent.

The patient population for the comparative valve study comprised 24 individuals; 42% were women. Additionally, 5 patients were recruited for a CMR reproducibility study. All patients completed the imaging protocols without difficulty. Other than obtaining informed consent, no additional patient preparation, instrumentation, or instruction was required over that of any CMR examination (eg, typically <5 minutes).

Imaging Protocol

Patients were imaged with CMR and echocardiography consecutively but in random starting order in the CMR suite during a single imaging session by experienced clinical technologists. Patients, in the supine position, were imaged sequentially, without repositioning, on the same MRI trolley table for both examinations. Both the sonographer and the CMR technologist acquired image and velocity data independently, without physician observation, and were blinded to the specific findings of the other. Echocardiographic data analysis was performed by the sonographer on acquisition and was used, unmodified, as the gold standard for comparison with CMR. The CMR data were compared directly with data acquired and analyzed independently by an experienced sonographer, blinded to the MRI data.

Echocardiographic Methods

With the use of an Acuson Sequoia imager with a 2.5-MHz, 128-element, phased-array transducer, images were acquired by using standard imaging windows with short breath-holds used as needed. Doppler flow data were acquired from the LV outflow tract (LVOT) region in pulsed wave mode and from the aortic valve in continuous wave mode in the 5-chamber view. The LVOT diameter was measured in the parasternal long-axis view in proximity to the position of the pulse wave Doppler data, then converted to LVOT area ($A_{LVOT}$) according to $A = \frac{pi}{4}r^2$. Peak velocities and VTIs, calculated with resident software at the time of imaging, were used to calculate pressure gradients according to the modified Bernoulli equations ($\Delta P = 4V^2$) and valve orifice areas according to the continuity equation approach [$A_{LVOT} = A_{LVOT}(VTI_{LVOT} \times VTI_{aortic})$]. Calculations were based on the single best representative heartbeat as

![Figure 1. For quantitative flow measurements, 5 views were planned for velocity-encoded flow imaging. Imaging planes, parallel to the aortic valve as seen in LVOT view, are positioned at the valve and on either side, with offsets of ±10 mm and ±15 mm.](Image 319x502 to 543x726)

selected independently by the expert sonographer, blinded to the MRI data.

CMR Methods

Patients were imaged through the use of a 1.5-T MRI scanner with a 5-element, phased-array coil (NT Intera CV R8, Philips Medical Systems). Initially, a standard ventricular function examination was performed by acquiring cine loops of standard views with the use of a steady-state gradient echo technique. These views provided not only qualitative assessment of function but also high-resolution and time-resolved images of the aortic valve plane. Quantitative images were obtained about the aortic valve plane by using a free-breathing, retrospectively gated velocity encoding technique sensitized for flow in the through-plane direction $\alpha = 6.0$ ms/2.9 ms/30°, 30 frames/heartbeat, 2 NSA, typical voxel size $= 1.0 \times 1.3 \times 9.0$ mm$^3$, depending on patient size). For a typical heart rate of 60 bpm, each scan takes 3.5 minutes. Typically, the maximum encoding velocity ($V_{ENC}$) was $\approx 2$ m/s for the LVOT and $\approx 4$ m/s in the aorta. For the aorta, $V_{ENC}$ was initially 2.5 m/s but was increased a priori if the peak velocity could be predicted to be greater, based on that patient’s functional images (eg, through valve motion or postvalve signal loss). Flow images were reacquired with correct $V_{ENC}$ if velocity aliasing occurred.

For the quantitative flow measurements, 5 views were planned by using the high-resolution cines from the functional examination for positioning imaging planes parallel to the aortic valve plane such that the middle imaging plane was positioned just to the aortic side of the valve when at its greatest excursion toward the apex (typically, end-systole). Four other planes were positioned parallel to this plane but offset $\pm 10$ mm and $\pm 15$ mm from its center (Figure 1.) The naming convention adopted for these slices is as follows, moving from the LVOT toward the aorta: Level−, Level−, Level−, Level+ and Level++. The data were transmitted to an offline image processing station (EasyVision R5, Philips Medical Systems), and the quantitative flow images were analyzed.

For each of the 5 CMR imaging levels, regions of interest (ROIs) were drawn on each of the 30 frames of the cine to include the LVOT, the aortic valve, or the aorta, depending on slice position. Flow data for the pixels within the ROI were exported to a spreadsheet, and the VTIs were calculated by using Simpson’s rule
to integrate, during systole, the peak flow velocity versus time curve (Figure 2).

The VTI was calculated for each of the 5 levels and compared with the corresponding Doppler ultrasound measurements of aortic VTI or LVOT VTI. By using the modified Bernoulli equation, peak and mean gradients were calculated and compared between the 2 modalities ($P_{\text{peak}} = 4(V_{\text{peak}})^2$ and $P_{\text{mean}} = 4(V_{\text{mean}})^2$). Aortic valve orifice dimension was calculated by applying the continuity equation, where the measured values are LVOT area, LVOT VTI, and aortic VTI; the ratio of LVOT and aortic VTIs is multiplied by the cross-sectional area of the LVOT.

Although numerous approaches exist to estimate the cross-sectional area of the LVOT by MRI, in ultrasound the practiced norm is to measure the diameter of the LVOT and assume the cross section to be circular. For consistency between modalities and ease of operation, the same approach was used to evaluate the MRI data. On the LVOT view at end systole (the same image used to plan the velocity-encoded images), the diameter of the LVOT was measured at the 2 positions corresponding to Level− and Level−− (ie, 10 mm and 15 mm from the valve plane, respectively). With 2 estimates of LVOT area and multiple VTI values calculated from the MRI flow data, estimations of the aortic aperture dimension were made with the use of data from all permutations of position.

To evaluate the consistency of the procedures developed for measuring valve size with velocity-encoded MRI, a reproducibility study was performed. Five patients underwent the CMR examination twice, and values from each examination were compared. Between consecutive MRI examinations, the patient was allowed to walk about, then repositioned in the MRI scanner. For the reproducibility study, the velocity-encoded data were limited to the Level− and Level−− positions, with LVOT diameter measured 10 mm from the valve plane, based on a posteriori knowledge that these provided the strongest correlation to ultrasound. All calculations were performed as above.

Statistics
For statistical analysis, the SAS System for Windows, version 8 (SAS Institute Inc) was used. Correlation between CMR and echocardiography methods was tested by regression analysis. A 2-tailed probability value <0.05 was considered evidence of statistical significance.

For evaluation of CMR reproducibility, intraclass correlation coefficients (ICC) were computed according to Shrout and Fleiss. To compute ICC, the general linear model procedure in SAS was used to perform ANOVA, which provided between- and within-subject mean square needed for the calculation of ICC. Bland-Altman analysis was also performed.

For more extensive error analysis of the VTI data, relations between echocardiography and CMR, in addition to linear regression, were evaluated using nonlinear equations that would best represent the measured data.

Results
The patient population (n=24), ranging from 34 to 84 years (mean, 58), exhibited a wide range in severity of aortic stenosis, with the aperture dimensions ranging from 0.5 to 1.8 cm², as measured by ultrasound. All patients were in sinus rhythm, and just more than half (54%) exhibited at least mild aortic regurgitation. The left ventricular ejection fraction averaged 63±10%, ranging from 40% to 80%. The calculations and comparisons of the individual AS measurements are presented below separated by flow and pressure parameters.

Analysis of the ROIs in the velocity-encoded MRI flow data provides detailed information regarding the flow profile across the aorta at the level of the imaging slice. The velocities measured by velocity encoding for individual voxels represent an average of all velocities present within that single 10 mm³ voxel at 30 different time points throughout the cardiac cycle. Figure 3 shows typical magnitude (A) and phase (B) images and a shaded surface mesh diagram (C) of one such flow profile at peak systole. In 5 of the patients,
the velocity-encoded image (at either Level+ or Level+++) had to be reacquired with a higher $V_{ENC}$ because of aliasing. The ROI analysis extracts the single greatest velocity from all included voxels, which is used as the measure of peak velocity for that ROI at that time and image plane level. As in ultrasound, this single maximal peak velocity can be used to estimate the peak pressure gradient. Similarly, the mean pressure gradient can be calculated. Considering the CMR measurements in the postvalve levels (where velocities reveal the pressure gradient across the valve), CMR data correlated well with ultrasound data for both peak and mean pressures: $r=0.82$ and 0.87, respectively, for Level+ and 0.82 and 0.83 for Level+++

VTI Comparisons

Perhaps more important than these “single value” estimates are the more robust velocity-time integral measurements, which include more data. Figure 2 (in Methods) shows a typical flow-velocity curve, with the shaded portion representing the VTI. The correlation between MR and ultrasound measures of VTI for both LVOT and aorta are plotted in Figure 4A.

Statistically, the Level−− from the MRI data correlated most strongly with the ultrasound measure of VTI in the LVOT and Level+++ with that in the aorta. However, there was no difference between the measurements at Level−− ($r=0.93$) or between Level+ and Level++ (r=0.96), suggesting relative insensitivity to distance of the imaging plane from the valve under these conditions.

The linear regression analysis of the LVOT VTI shown in Figure 4A suggests a relation between MRI and ultrasound, with a slope of 0.86 and an intercept of virtually zero. For the aortic VTI data, the slope and intercept for the linear regression analysis are 0.63 and 0.22, respectively. Pooling all VTI data (aortic and LVOT) together, a linear fit describes the data with slope and intercept of 0.82 and 0.08 ($r=0.95$). It is apparent that a modest underestimation of the severity of aortic valve stenosis might occur with velocity-encoded CMR for VTIs greater than $\sim0.8$ m (Figure 4B). Limiting the data

Figure 3. From velocity-encoded MRI, both magnitude (A) and phase (B) images can be reconstructed. Phase image encodes measured velocities at each pixel location. In this example, near peak systole, the aorta has been isolated and displayed as a shaded surface plot (C) of the velocity profile. Note incomplete opening of the aortic valve (arrow).

Figure 4. VTI measurements made by MRI correlate well with those made by Doppler ultrasound. Separated by LVOT or aorta (A), data are described well with linear models. Considering all data combined (B), an exponential model fits the data better ($r=0.96$). For VTIs greater than $\sim0.8$ m, there may be a modest underestimate of VTI by MRI (shaded area). Restricting linear regression to all data $<0.8$ m (by ultrasound), the resulting relation between MRI and ultrasound is virtually that of identity ($r=0.97$).
set to all VTIs $<0.8 \text{ m}$, as measured by ultrasound, the linear regression yields a slope of 1 and an intercept of 0 ($r=0.97$), which is consistent with the hypothesis that CMR could underestimate the aortic VTI for the most severe cases of aortic stenosis. If one fits an exponential equation to all the data, one gets $y=1.39(1-e^{-0.94x})$ as the fit ($r=0.96$), which describes the data at low VTI as well as the higher values. The shaded area between the exponential and linear fits in Figure 4B represents the potential error of CMR in underestimating aortic VTI at these high values.

Valve Area Comparisons

With the measurement of 2 LVOT areas and 2 values of VTI for each of the LVOT and the aorta, 8 permutations of the equation to calculate the valve size exist. The correlation analyses for these 8 calculations are presented in Table 1, as compared with the Doppler ultrasound.

Although most of the combinations of CMR data provide valve size measures corresponding well to ultrasound, the valve areas computed by using the LVOT diameter measurement from the Level– (1 cm from the valve plane) are better for selected combinations of data. However, calculations derived by using the different slice positions for VTI produced similar outcomes in terms of acceptable correlation figures. The single best correlation between CMR and ultrasound data ($r=0.83$, $P<0.001$) was between values derived by using the VTI measurements at the Level++ and Level– and the diameter measurement closer to the valve at Level–. Figure 5A depicts the valve sizes derived from this combination of CMR data plotted versus ultrasound for the complete population. The Bland-Altman plot is shown in Figure 5B, indicating that the methods agree, exhibiting a mean difference near zero and a spread within 2 SD. Valve sizes were also calculated by relaxing the assumption that the LVOT cross section is circular. By using planimetry on the velocity-encoded images (magnitude reconstruction) at Level– or Level– to obtain LVOT area, calculated valve size was statistically greater than with ultrasound (which did not provide planimetric LVOT area), with corresponding lower correlation coefficients ($r=0.69$, $P<0.001$ for both). By using the approach with only Level++ and Level– and estimating LVOT area by diameter, the average (±SD) total time required to analyze the velocity-encoded image data to obtain VTI and valve size measurement was $3.9\pm1.3$ minutes per patient.

Reproducibility of CMR

For the 5 patients receiving duplicate MRI examinations, the data were analyzed as above. The pressure gradient measurements, both mean and peak, correlate well. Likewise, all other repetitive measurements were similar. Table 2 summarizes the comparison of all the duplicate measurements with both the Pearson correlation coefficient ($r$) and the more rigorous intraclass correlation coefficient (ICC). The repeat calculation of valve size, the end result of all the measurements, are compared in Figure 6, with $r=0.94$ ($P<0.05$) and ICC=0.87.

Discussion

A majority of transthoracic echocardiograms in a busy clinical service entails the assessment of valve disease. The real-time capabilities of echocardiography, together with its
accuracy, rapid quantitative bedside image postprocessing, and reasonable cost, render it an ideal method for such screening assessments. Echocardiography and CMR, providing many similar diagnostic capabilities, could be considered interchangeable for certain requests. Yet, the economics of routine, widespread clinical adoption of CMR mandate that it serve as a replacement test for echocardiography, covering the full functionality, thereby avoiding unnecessary incremental costs. To this end, an optimized CMR approach for screening and quantitative analysis of valve disease appears necessary. Although CMR provides exquisite documentation of function, chamber dimensions, and ventricular mass, the challenge is to develop a rapid, reliable, and repeatable method for quantitative valve assessment that is an acceptable alternative to echocardiography.

Accordingly, the present work was conducted to establish a practical “cookbook” approach for quantification of aortic valve stenosis that could be carried out independently by CMR technologists without direct physician supervision. Kilner et al\textsuperscript{16} were among the first to show in 2 patients the potential for quantitative assessment of aortic valve stenosis by CMR. Soon thereafter, Sondergaard et al\textsuperscript{5,17} used velocity-encoded CMR to estimate orifice area (through planimetry of the peak systolic velocity jet) in 12 patients, illustrating good correlation with invasive catheterization. Other investigators also have demonstrated excellent correlations between CMR and invasive hemodynamic data for assessment of instantaneous peak aortic valve velocities and pressure gradients.\textsuperscript{8,9} However, optimized and robust CMR protocols for these purposes heretofore have not been defined and compared with echocardiographic methods such as the VTI/continuity equation approach.

Doppler ultrasound recordings of valve velocity are acquired in a direction parallel to flow, which requires that the sonographer search for the peak velocity envelope dynamically by manipulating the transducer. Visual, aural, and stereotactic cues combine to make this a rapid process that minimizes time to convergence to acceptable data. Dynamic adjustment of Doppler parameters also speeds the process. However, poor echocardiographic windows may compromise recording quality, and unusual anatomic configurations (ecstatic aortas, horizontal heart positions) may preclude exact parallel orientation of the Doppler beam with the high-velocity aortic jets, mandating recording from other approaches such as the suprasternal notch.

In contrast, the CMR imaging planes generally are specified a priori, parallel to the plane of the aortic valve, and velocity encoding is typically selected only in the through-plane direction to reduce data acquisition duration. This technique assumes that flow is perpendicular to the valve (image) plane. If this is not the case, an undersampling of the peak velocity can occur, which depends on the angle between the flow jet and the imaging plane. The most rigorous method to measure the velocity would be to acquire complete 3D velocity encoding data, at the expense of increased imaging time and patient discomfort.

Thus, CMR typically is restricted to preformatted acquisition of the data, followed by post hoc review. To examine this potential limitation, we retrospectively quantified the effect

### Table 2. Statistical Comparisons Between Repeat Measurements of Velocities, Pressure Gradients, and VTI With Velocity-Encoded CMR

<table>
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<th>LVOT (Level − −)</th>
<th>Aorta (Level+ +)</th>
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<tr>
<td></td>
<td>r</td>
<td>ICC</td>
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<tr>
<td>Peak velocity, m/s</td>
<td>0.80</td>
<td>0.80</td>
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<tr>
<td>Peak gradient, mm Hg</td>
<td>0.85</td>
<td>0.84</td>
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<tr>
<td>Mean gradient, mm Hg</td>
<td>0.91†</td>
<td>0.86</td>
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<tr>
<td>VTI, m</td>
<td>0.91†</td>
<td>0.86</td>
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*P < 0.01.
†P < 0.05.
of misalignments between imaging planes and flow jet on our velocity measurements in 8 patients. The angle between the actual velocity-encoded imaging plane and the flow jet was determined from the LVOT view, and velocities were corrected by the cosine of this angle. The errors (ie, differences) estimated were all <6% (mean 3%), which is consistent with similar error analysis done by Firmin et al.22 Thus, in general, for clinical applications of such “cookbook” approaches, the error in image plane orientation is expected to be modest.

Opportunities for other errors and inefficiency arise in the spatial orientation of the imaging plane, its proximity to the valve, and in the range of values chosen for velocity encoding (V\textsubscript{ENC}). Important differences also exist in the physics of signal acquisition and in signal processing. Continuous wave Doppler samples the velocities in a cylindrical volume element (voxel) of space. A velocity envelope is recorded that registers all Doppler shifts (ie, all velocities) within that voxel. Typically, for laminar flow in the center of the jet, a well-defined velocity envelope exists, with a preponderance of values clustered around some peak velocity level. The sonographer can easily trace this envelope to perform immediate computer-assisted planimetry of the VTIs. Thus, Doppler echocardiography registers peak velocities within a small voxel located in the center of the jet.

Velocity-encoded CMR registers the average velocity within a single imaging voxel, but many velocity values are depicted across the extent of the jet as it intersects the imaging plane. The voxel size is specified by the field of view, the size of the sampling matrix for in-plane resolution, and the thickness of the imaging plane for through-plane resolution. Typical voxel dimensions are 1×1×10 mm\textsuperscript{3} for velocity imaging. A number of other phenomena affect CMR velocity data, such as eddy currents and ghosting artifacts, but for clinical purposes are generally ignored or only roughly corrected. Furthermore, the fact that the aortic valve moves during contraction relative to the stationary imaging plane imposes another consideration often overlooked. Kozerske et al\textsuperscript{26} suggest an elegant method of slice following to correct for this, but this technique is not widely available clinically.

Regardless of these potential sources of difference, it is remarkable that the VTI data are so similar (regression slopes and correlation coefficients approaching unity), with no apparent systematic offsets (Y-intercepts ~0). The notable exception to this linear relation exists in patients with severe aortic stenosis in whom the aortic VTI exceeds 0.8 m. The extent of this error is found in the shaded area between linear and exponential curve fits in Figure 4B. The lower ranges in this plot demonstrate an excellent linear relation between methods. The upper range where CMR and echocardiographic data begin to diverge, which can result in underestimation of valve stenosis by CMR, occurs beyond the threshold for defining critical stenosis. This point is reached when the ratio of aortic to LVOT VTIs approach 4:1, for normal LV function. For patients with abnormal LV function and severe aortic stenosis, VTI values can be lower and hence the threshold reduced, but CMR should perform well under those conditions.

The choice of CMR imaging plane position may not be crucial, within the limits of the 1 to 1.5 cm from the valve. Therefore, extreme fastidiousness is not required to position the imaging planes for calculation of VTI, within the bounds established in this study. Likewise, for defining regions of interest within the aorta and outflow tract, only modest effort is required to ensure inclusion of the total flow jet, but laborious manual tracing around the vessel wall or flow jet is not needed. Thus, the proposed technique appears to be user-friendly for both image acquisition and postprocessing analysis, much like the case for echocardiography.

The continuity equation approach with the use of VTI data are adapted from current echocardiographic methods and appears to be both robust and easy to perform. Velocity-encoded image acquisition and postprocessing is easily implemented on most clinical scanners. Compared with echocardiography, the examination time would be roughly similar—well less than 20 minutes. Acquiring the 2 velocity images would add just over 6 minutes to a standard CMR functional examination, also about 6 minutes long.27 The associated postprocessing, as described, would add another few minutes, less if automated. Patient preparation (ie, ECG electrode and surface coil placement) is virtually the same as for echocardiography and generally <5 minutes. Perhaps the most unforgiving parameter is the V\textsubscript{ENC}. In acquiring the flow data, care must be taken to avoid velocity aliasing, which occurred in only 3 of 24 patients when evaluated by the optimal “cookbook” approach. Until real-time flow recording becomes available, adjustments of this parameter post hoc might continue to represent a major source of inefficiency.

In conclusion, CMR offers high spatial resolution analysis of cardiac anatomy and myocardial function. Additionally, the implementation of velocity-encoded MRI provides a tool for implementing a user-friendly approach for analyzing stenotic aortic valves with a robust and simple technique requiring minimal postprocessing. Such routine methods should contribute to the clinical adoption of CMR as an alternative to echocardiography or catheterization in selected cases.

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References

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