Cardiovascular Magnetic Resonance Imaging for Valvular Heart Disease

Technique and Validation

Peter J. Cawley, MD; Jeffrey H. Maki, MD, PhD; Catherine M. Otto, MD

Over the last half century, clinicians have employed several means to advance our knowledge of the causes and consequences of valvular heart disease. Invasive cardiac catheterization provided valuable information about hemodynamics, 2-dimensional (2D) echocardiography (echo) allowed direct visualization of the valvular apparatus and cardiac chambers, and Doppler echocardiography afforded a noninvasive tool for assessing hemodynamics and disease severity.

Echocardiography is now the standard tool for initial assessment and longitudinal evaluation of patients with valvular heart disease; however, echocardiography is limited in patients with poor acoustic windows and may be more operator dependent than other modalities, particularly for quantitation of disease severity. In the last 20 years, cardiovascular magnetic resonance (CMR) has emerged as an alternative noninvasive modality without ionizing radiation that is applicable to patients with valvular heart disease. CMR provides images of valve anatomy and allows quantitative evaluation of stenosis and regurgitation. CMR can also discern the consequences of the valvular lesion, including the effects of ventricular volume or pressure overload and alterations in systolic function. The purpose of the present review is to summarize the general principles of CMR and validate CMR as a tool for evaluation of valvular heart disease.

General Principles

CMR uses a variety of pulse sequences to assess valvular heart disease (Table 1). A pulse sequence is a combination of transmitted radiofrequency pulses and magnetic gradients in the presence of a strong external magnetic field, from which a series of received radiofrequency pulses or “echoes” are obtained and processed into an image.1,2

Anatomy

CMR has the potential to visualize all parts of the valve (leaflets, chordae tendineae, and papillary muscles) throughout the entire cardiac cycle. Congenitally abnormal valve leaflets (bicuspid), aberrant papillary muscles or aberrant chordal attachments (parachute mitral valve), leaflet thickening, presence and extent of calcification, leaflet redundancy and prolapse, and commissural fusion are all anatomic descriptions that have been reported by CMR. However, 2-dimensional (2D) echocardiography is superior for imaging structures that are thin and highly mobile owing to its greater temporal resolution and the absence of partial volume effects. Although 2D echocardiography remains the primary approach for visualization of valve anatomy, CMR is a reasonable alternative if ultrasound windows are poor.

In addition, CMR can provide visualization of valve masses such as vegetations, thrombi, or tumors, including attachment site and mobility.3–6 For masses of sufficient size, tissue characterization may be helpful when the origin of the mass is unknown.4,6 However, the minimum size of the mass needed for detection by CMR has not been described, and sensitivity and specificity compared with 2D echocardiography have not been evaluated.

The steady-state free precession (SSFP) cine pulse sequence is the most widely used CMR pulse sequence for assessing valve anatomy and motion (Figures 1 and 2). This pulse sequence has excellent blood-to-myocardium contrast and a high intrinsic signal-to-noise ratio and has largely replaced gradient echo as the preferred pulse sequence for cine imaging of valve anatomy. SSFP produces a 2D image in any prescribed plane having multiple phases (frames) throughout the cardiac cycle, with a typical temporal resolution of 25 to 50 ms. To produce an SSFP cine image throughout all of systole and diastole, image acquisition is gated to the ECG and occurs over several cardiac cycles, easily obtained in a single breath hold (6 to 12 seconds). Non-cine pulse sequences such as turbo spin echo (T1 weighted, T2 weighted, fat saturation) and segmented inversion recovery gradient recalled echo pulse sequences may aid in tissue characterization of valve masses.4,6

Velocity

SSFP and gradient echo cine pulse sequences can visualize flow turbulence (Figure 3) on the basis of loss of signal...
due to the dephasing of moving protons. This approach can help show the location of "jets" and optimize the location of velocity sampling. Although SSFP provides improved visualization of valve anatomy, it is less sensitive for depicting flow disturbances. This can lead to underestimation or the overlooking of a regurgitant jet. Gradient echo cine pulse sequences, on the other hand, are more sensitive for the detection and sizing of regurgitant jets. With gradient echo sequences, the sensitivity for detecting dephasing (eg, valve regurgitation) is a function of the echo time: the longer the echo time, the larger and more pronounced the regurgitant jet.7

Phase-contrast pulse sequences (other names include velocity-encoded cine, Q flow, or velocity mapping) are used for velocity measurements and are based on the accumulated phase of moving protons. In this pulse sequence, bipolar gradients oriented in the expected direction of blood flow are applied to each frame of the imaging slice of interest to induce phase shifts.8 Phase refers to the angular position of an individual proton’s spin vector with respect to a frame of reference. Stationary objects within this slice have a net phase of zero, because all phase induced by the first lobe of the bipolar gradient is reversed by the second lobe. Moving objects (blood) gain a net phase depending on the direction of blood flow, and this net phase (or phase shift) is proportional to the velocity of blood. This net phase can be displayed as a phase map with differences in signal intensity representing different velocities (Figure 4). Pixels depicting flow in the phase-encoding direction appear bright (Figure 4A), and flow opposite to the phase-encoding direction appears dark (Figure 4B). Objects with a phase shift of zero (stationary) are gray or speckled, as can be seen in the lungs or chest wall.

Velocity mapping requires that the appropriate maximum velocity be programmed into the pulse sequence. Aliasing occurs if the angular phase shift is >180°, and the velocity within that pixel is then misregistered. This occurs if the programmed maximum velocity is less than the sampled velocities of blood flow in the imaging slice. The closer the programmed maximum velocity is to the maximum velocity present, the greater the sensitivity and accuracy of this technique to detect lower velocities within the region of interest.

Velocity mapping produces 2 sets of images: magnitude image and phase velocity maps (Figure 4). The magnitude image is used for anatomic orientation of the imaging slice and to identify the boundaries of the vessel imaged. Blood has an increased signal, whereas turbulent flow is depicted with signal loss within the magnitude image. The phase map encodes the velocities within each pixel. Using both images, a region of interest can be traced on each time frame of the data set. The region of interest must be drawn for each frame of the cardiac cycle carefully because of movement and deformation of the vessel. Within each region of interest, the peak instantaneous velocity (V_{max}) for each time frame can also be obtained. With the simplified Bernoulli equation (4V^2_{max}), the peak instantaneous gradient can be estimated by substituting in the peak instantaneous velocity. Mean pressure gradients are obtained by averaging all of the instantaneous velocities over systole. With flow phantoms, phase velocity mapping has been shown to accurately measure velocities over 5 m/s.9

Flow Volume
Velocity maps can also be used to determine flow volume throughout the cardiac cycle. With the same magnitude and
phase velocity maps, a region of interest is traced around the vessel lumen to determine the area of the vessel, frame by frame. By multiplying the velocity (cm/s) of each pixel by the area (cm²) of the region of interest, the instantaneous flow volume (cm³/s) is obtained for each frame of the cardiac cycle. The instantaneous flow volume of each frame (y-axis) can be plotted against the time of the cardiac cycle (x-axis) to show bulk flow as it relates to the cardiac cycle (Figure 5). When the area under the curve is integrated for systole and diastole, forward and regurgitant volumes can be generated.

Stroke-volume measurements by velocity mapping in the ascending aorta have been shown to have a strong correlation to in vivo stroke-volume measurements by cine CMR pulse sequences of the left ventricle in subjects without significant mitral regurgitation, in vivo stroke volume by the Fick principle and by thermodilution, in vivo stroke-volume measurements by Doppler, and in vitro stroke-volume measurements within continual and pulsatile flow phantoms.10–14

Potential Pitfalls for Phase-Contrast Imaging
Although the “in-plane” pixel size for phase-contrast imaging may be on the order of 1.5 to 2.0 mm, the slice thickness (volume of tissue) is typically 6 to 8 mm. Thus, vena contracta velocities may be underestimated owing to averaging of flow velocities from inside and outside the actual vena contracta (partial volume averaging). In these circumstances, velocity measurements for each pixel may be lower than actual velocities. Furthermore, the smaller the structure of interest, the fewer pixels are fully within this region, which

Figure 2. A, Systolic frame of a 3-chamber SSFP image of myxomatous mitral valve disease. Note that the aortic valve leaflets are opened. The anterior and posterior mitral valve leaflets are redundant. B, Two phases later in systole, significant prolapse of both the anterior and posterior mitral valve leaflets occurs. The dashed arrow represents signal void from mitral regurgitation. C, One phase later in systole, mitral regurgitation worsens (dashed arrow). This is an example of late systolic mitral regurgitation. This patient had a mitral regurgitant fraction of 40%. Involvement of all segments of both mitral valve leaflets suggests a low likelihood of valve repairability. Ao indicates aorta; LV, left ventricle; and PM, papillary muscle.

Figure 3. A, Three-chamber SSFP image of the left ventricular outflow tract. In systole, the arrow demonstrates flow turbulence (signal void) that begins at the level of the aortic valve leaflets. This patient has an antegrade velocity of 3.1 m/s. In diastole, the arrow demonstrates flow turbulence (signal void) that begins at the level of the aortic valve, consistent with central aortic regurgitation. This patient had a regurgitant fraction of 39%. B, SSFP image of the right ventricular outflow tract in a patient with repaired tetralogy of Fallot. Essentially no pulmonic valve leaflets remain, and pulmonic regurgitation is wide open (arrows demonstrate signal void), with a regurgitant fraction of 55%. LA indicates left atrium; LV, left ventricle; RA, right atrium; and RV, right ventricle.
amplifies the effects of partial volume averaging. Limited temporal resolution reduces the accuracy of CMR velocity measurements. Lower frame rates may not be able to capture high velocities of short duration, which results in underestimation of peak velocities. In addition, it is very important for the imaging slice to be oriented perpendicular to the flow of blood. If the angle of intercept is not 90°, an increased likelihood exists of inaccurate velocity measurements.

Ventricular Volumes and Function

Left ventricular volume assessment by CMR has been performed in several ways. In 1 method, the same formula used for biplane left ventriculography can be applied to the 2- and 4-chamber CMR views. However, these types of formulas depend on left ventricular geometric assumptions that may not apply to any individual patient. Left and right ventricular volume assessment is optimally performed with multislice 2D SSFP cine imaging covering both ventricles. A parallel stack of serial images (short-axis or 4-chamber) is acquired from base to apex of the left ventricle with a 6- to 8-mm slice thickness. Most commonly, an imaging "gap" of 2 to 4 mm is used to balance the accurate assessment of ventricular volumes against excessive prolongation of total scan time. Volume for each image plane is calculated as the area of the endocardial tracing multiplied by the addition of the image slice thickness and interslice gap. End-diastolic and -systolic volumes are calculated by summing all slices, which allows for calculation of stroke volume, cardiac output, and ejection fraction with standard equations.

The accuracy of CMR-calculated LV volumes has been validated both in vitro and in vivo. In vitro, a close correlation has been shown between CMR ventricular volumes and fluid-filled phantoms, wax casts of porcine ventricular cavities, latex casts of cadaveric hearts, and dynamic (pulsatile-flow) phantoms. In vivo, a close correlation between left and right ventricular stroke volumes has been shown in normal subjects with excellent reproducibility. Right ventricular volumes have also been validated in vitro to casts of cadaveric hearts.

Areas of delayed gadolinium enhancement of the left ventricular myocardium correlate histopathologically with infarction but have also been described in patients with hypertrophic, dilated, and infiltrative cardiomyopathies. Several patterns of delayed gadolinium enhancement have been reported in adults with severe aortic stenosis, although the clinical implications of these findings in terms of prognosis and timing of intervention have not been fully elucidated.

Evaluation of Valve Stenosis

Standard measures of stenosis severity are used for CMR: peak antegrade velocity, pressure gradient, and valve area.
Peak Antegrade Velocity and Pressure Gradient

Validation studies in human subjects have been published as early as the 1990s showing that antegrade velocity correlates well with continuous-wave Doppler echocardiography in adults with aortic stenosis. A trend can be identified for CMR to underestimate the peak velocity, most likely because of partial volume averaging within the vena contracta (Table 2).23–31 In patients with mitral stenosis, inflow velocities as assessed by Doppler echocardiography and CMR phase-contrast imaging correlate well (Table 3).23,32–37

Table 2. Select Studies Validating Indices of Aortic Stenosis by CMR

<table>
<thead>
<tr>
<th>First Author (Year)</th>
<th>Principle</th>
<th>Reference</th>
<th>n</th>
<th>r</th>
<th>Mean Difference ± 1 SD (CMR–Echo)</th>
<th>CMR Reproducibility*: Mean Difference ± 1 SD</th>
</tr>
</thead>
<tbody>
<tr>
<td>Kilner23 (1993)</td>
<td>$V_{\text{max}}$ peak $\Delta P$</td>
<td>TTE</td>
<td>26†</td>
<td>...</td>
<td>$-0.10 \pm 0.46 \text{ m/s}$</td>
<td>$0.11 \pm 0.29 \text{ m/s}$‡</td>
</tr>
<tr>
<td>Eichenberger24 (1993)</td>
<td>peak $\Delta P$</td>
<td>TTE</td>
<td>15</td>
<td>...</td>
<td>$2.6 \pm 13.3 \text{ mm Hg}$</td>
<td>...</td>
</tr>
<tr>
<td>Sondergaard25 (1993)</td>
<td>$V_{\text{max}}$ mean $\Delta P$</td>
<td>TTE</td>
<td>12</td>
<td>...</td>
<td>$-0.68 \pm 0.91 \text{ m/s}$</td>
<td>...</td>
</tr>
<tr>
<td>Caruthers26 (2003)</td>
<td>peak $\Delta P$</td>
<td>TTE</td>
<td>24</td>
<td>0.82</td>
<td>...</td>
<td>$r=0.94$§</td>
</tr>
<tr>
<td>Caruthers26 (2003)</td>
<td>mean $\Delta P$</td>
<td>TEE</td>
<td>24</td>
<td>0.83</td>
<td>...</td>
<td>$r=0.94$§</td>
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<td>John27 (2003)</td>
<td>planimetry</td>
<td>TEE</td>
<td>40</td>
<td>0.96</td>
<td>$0.02 \pm 0.08 \text{ cm}^2$‡</td>
<td>$0.07 \pm 0.06 \text{ cm}^2$‡</td>
</tr>
<tr>
<td>Kupfahl28 (2004)</td>
<td>planimetry</td>
<td>TEE</td>
<td>32</td>
<td>...</td>
<td>$0.02 \pm 0.21 \text{ cm}^2$‡</td>
<td>$0.03 \pm 0.05 \text{ cm}^2$‡</td>
</tr>
<tr>
<td>Debl29 (2005)</td>
<td>planimetry</td>
<td>TEE</td>
<td>25</td>
<td>0.86</td>
<td>$0.13 \pm 0.16 \text{ cm}^2$‡</td>
<td>...</td>
</tr>
<tr>
<td>Reant30 (2006)</td>
<td>planimetry</td>
<td>TEE</td>
<td>39</td>
<td>0.58</td>
<td>$0.01 \pm 0.14 \text{ cm}^2$‡</td>
<td>$0.03 \pm 0.14 \text{ cm}^2$‡</td>
</tr>
<tr>
<td>Schlosser31 (2007)</td>
<td>planimetry</td>
<td>TEE</td>
<td>32</td>
<td>0.82</td>
<td>$0.15 \pm 0.13 \text{ cm}^2$‡</td>
<td>$0.75$**‡</td>
</tr>
</tbody>
</table>

n Indicates sample size; TTE, transthoracic echocardiography; peak $\Delta P$, peak pressure gradient; mean $\Delta P$, mean pressure gradient; and TEE, transesophageal echocardiography.

If not stated in the publication, statistics were calculated from the data provided in the manuscript. If this was not performed or data were not provided, a designation of “not available” (...) was given.

*Reported as mean difference ± 1 SD; if not available, the statistical test provided by the author is stated.
†This analysis included 17 aortic stenosis and 9 mitral stenosis measurements.
‡Interobserver reproducibility.
§Interstudy reproducibility.
| | Cine MRI pulse sequence was gradient echo; the other planimetry studies used an SSFP cine sequence.
¶Mean absolute difference.
#Intraobserver reproducibility.
**Kendall’s statistic.
Table 3. Select Studies Validating Indices of Mitral Stenosis by CMR

<table>
<thead>
<tr>
<th>First Author (Year)</th>
<th>CMR Method</th>
<th>Reference Standard: Method</th>
<th>n</th>
<th>r</th>
<th>Mean Difference±1 SD (CMR–Other Modality)</th>
<th>CMR Reproducibility*: Mean Difference±1 SD</th>
</tr>
</thead>
<tbody>
<tr>
<td>Velocity/gradients</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Mohiaddin42 (1991)</td>
<td>V_{max}</td>
<td>TTE: V_{max}</td>
<td>5</td>
<td>...</td>
<td>−0.12±0.27 m/s</td>
<td>...</td>
</tr>
<tr>
<td>Kilner43 (1993)</td>
<td>V_{max}</td>
<td>TTE: V_{max}</td>
<td>26</td>
<td>...</td>
<td>0.10±0.46 m/s</td>
<td>0.11±0.29 m/s†</td>
</tr>
<tr>
<td>Hartiaa35 (1993)</td>
<td>E velocity</td>
<td>TTE: E velocity</td>
<td>10</td>
<td>0.68</td>
<td>...</td>
<td>0.16%§</td>
</tr>
<tr>
<td>Heidenreich36 (1995)</td>
<td>peak ΔP</td>
<td>TTE: peak ΔP</td>
<td>14</td>
<td>0.89</td>
<td>Vmax: 0.38±0.2 m/s</td>
<td>0.001 m/s‡</td>
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<tr>
<td>Valve areas</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Lin37 (2004)</td>
<td>T½, T2</td>
<td>TTE: T½</td>
<td>17</td>
<td>0.86</td>
<td>0.5±0.59 cm²</td>
<td>r=0.96‡</td>
</tr>
<tr>
<td>Djavidani38 (2005)</td>
<td>planimetry</td>
<td>TTE: T½</td>
<td>22</td>
<td>0.81</td>
<td>0.13±0.24 cm²</td>
<td>0.03±0.01 cm²†¶</td>
</tr>
<tr>
<td></td>
<td>catheterization: Gorlin</td>
<td></td>
<td>17</td>
<td>0.89</td>
<td>0.08±0.22 cm²</td>
<td>0.04±0.02 cm²¶</td>
</tr>
<tr>
<td>Djavidani39 (2006)</td>
<td>planimetry**</td>
<td>TTE: T½</td>
<td>13</td>
<td>0.98</td>
<td>0.03±0.09 cm²</td>
<td>...</td>
</tr>
<tr>
<td></td>
<td>catheterization: Gorlin</td>
<td></td>
<td>13</td>
<td>0.95</td>
<td>0.13±0.15 cm²</td>
<td></td>
</tr>
</tbody>
</table>

n Indicates sample size; TTE, transthoracic echocardiography; and T½, pressure half time.

If not stated in the publication, statistics were calculated from the data provided in the manuscript. If this was not performed or data were not provided, a designation of "not available" (...) was given.

*Reported as mean difference±1 SD; if not available, the statistical test provided by the author is stated.
†This analysis included 17 aortic stenosis and 9 mitral valve measurements.
‡Interobserver reproducibility.
¶Coefficient of variation.
§Normal volunteers only. Interobserver reproducibility: (observer 1−observer 2)/(mean of observer 1 and observer 2) and expressed in percentages.
| Valve Area           | In aortic stenosis, the valve area can be defined in 2 different ways: anatomic valve area and physiological valve area. An anatomic valve area is defined as the planimetered area of maximal opening of the aortic valve leaflets. CMR planimetry for the assessment of anatomic aortic valve area is based on direct visualization of the valve orifice, made possible by the excellent blood-to-myocardium contrast and high signal-to-noise ratio provided by SSFP (Table 2).23–31 SSFP planimetry has been shown to correlate better with planimetry using |

Table 4. Select Studies Validating Methods of Aortic Regurgitant Severity by CMR

<table>
<thead>
<tr>
<th>First Author (Year)</th>
<th>CMR Method</th>
<th>Reference Standard: Method</th>
<th>n</th>
<th>r</th>
<th>CMR Reproducibility</th>
</tr>
</thead>
<tbody>
<tr>
<td>Globits (1990)45</td>
<td>spin echo cine*: RF with biventricular volumes</td>
<td>catheterization: RF</td>
<td>20</td>
<td>0.91</td>
<td>interobserver mean variation in RVS=7 mL, LVSV=7.3 mL</td>
</tr>
<tr>
<td>Aurigemma46 (1991)</td>
<td>gradient echo cine: regurgitant jet area by signal void</td>
<td>TTE: regurgitant jet area by color Doppler</td>
<td>13</td>
<td>0.88</td>
<td>...</td>
</tr>
<tr>
<td>Dulce47 (1992)</td>
<td>velocity mapping: RV, RF</td>
<td>gradient echo cine: RV and RF with biventricular volumes</td>
<td>10</td>
<td>RV 0.975; RF 0.991</td>
<td>interstudy: RV r=0.977; RF r=0.986; interobserver: RV r=0.99, RF r=0.99</td>
</tr>
<tr>
<td>Honda48 (1993)</td>
<td>velocity mapping: RF</td>
<td>TTE: regurgitant area by color Doppler</td>
<td>26</td>
<td>...</td>
<td>interobserver r=0.956</td>
</tr>
<tr>
<td></td>
<td>catheterization: angiographic grade by aortogram</td>
<td></td>
<td>9</td>
<td>0.8</td>
<td>intraobserver r=0.998</td>
</tr>
<tr>
<td>Sondergaard49 (1993)</td>
<td>velocity mapping: RF</td>
<td>catheterization: angiographic grade by aortogram</td>
<td>9</td>
<td>0.8</td>
<td>...</td>
</tr>
<tr>
<td>Levy50 (2007)</td>
<td>velocity mapping: RF</td>
<td>TTE: pulsed Doppler RF</td>
<td>30</td>
<td>0.68</td>
<td>...</td>
</tr>
</tbody>
</table>

n Indicates sample size; RF, regurgitant fraction; interobserver, interobserver reproducibility; interstudy, interstudy reproducibility; RVS, right ventricular stroke volume; LVSV, left ventricular stroke volume; TTE, transthoracic echocardiography; RV, regurgitant volume; and intraobserver, intraobserver reproducibility.

If not stated in the publication, statistics were calculated from the data provided in the manuscript. If this was not performed or data were not provided, a designation of "not available" (...) was given.

*Regurgitant fraction (%) = RVS−RVS/LVSV×100.
†RV=LVSV−RVS; RF (%) = (LVSV−RVS)/LVSV×100.
‡RF (%) = total stroke volume−forward stroke volume/total stroke volume×100.
transesophageal echocardiography than does gradient echo cine imaging.\textsuperscript{31} Some of these studies have included patients with bicuspid aortic valves. Planimetry, however, is a less than optimal approach in patients with calcific stenosis. This is because leaflet calcification and jet turbulence can make accurate visualization of the true orifice difficult and because of the complex, 3-dimensional shape of the stenotic orifice. Tomographic measurements assume a planar orifice that lies entirely within the image plane. In addition, physiological valve area, which correlates better with clinical outcome, is smaller than anatomic valve area owing to contraction of the flow stream as it passes through the narrowed orifice.\textsuperscript{38}

Few data are available comparing physiological (continuity equation) valve area between CMR and echocardiography, although this approach is feasible on the basis of CMR velocity mapping of the velocity-time integral in the left ventricular outflow tract and aortic valve orifice.\textsuperscript{26} The different concepts underlying anatomic and physiological valve areas may explain some of the apparent discrepancies in comparisons of diagnostic approaches. These comparisons are most appropriate when measurements based on similar concepts are compared (Tables 2 and 3).\textsuperscript{23–37}

For evaluation of mitral stenosis, 2 methods are commonly used to determine valve area: planimetry and pressure half-time.\textsuperscript{39} A trend can be identified for CMR to overestimate the

### Table 5. Select Studies Validating Methods of Mitral Regurgitant Severity by CMR

<table>
<thead>
<tr>
<th>First Author (Year)</th>
<th>CMR Method</th>
<th>Reference Standard: Method</th>
<th>n</th>
<th>r</th>
<th>CMR Reproducibility*: Mean Difference±1 SD</th>
</tr>
</thead>
<tbody>
<tr>
<td>Sechtem\textsuperscript{51} (1988)</td>
<td>gradient echo cine\textsuperscript{†}: RF with biventricular volumes</td>
<td>TTE\textsuperscript{‡}: pulsed Doppler RF</td>
<td>8</td>
<td>0.9</td>
<td>r=0.96$; r=0.99$</td>
</tr>
<tr>
<td>Nishimura\textsuperscript{52} (1989)</td>
<td>gradient echo cine: regurgitant jet area and length by signal void</td>
<td>TTE: jet area</td>
<td>20</td>
<td>0.71</td>
<td>r=0.96$</td>
</tr>
<tr>
<td>Glogar\textsuperscript{53} (1989)</td>
<td>spin echo cine\textsuperscript‡: RF with biventricular volumes</td>
<td>catheterization: RF</td>
<td>13</td>
<td>0.84</td>
<td>...</td>
</tr>
<tr>
<td>Aurigemma\textsuperscript{54} (1990)</td>
<td>gradient echo cine: regurgitant jet area by signal void</td>
<td>catheterization: angiographic grade by aortogram</td>
<td>20</td>
<td>0.73</td>
<td>...</td>
</tr>
<tr>
<td>Glogits\textsuperscript{45} (1990)</td>
<td>spin echo cine#: RF with biventricular volumes</td>
<td>catheterization: RF</td>
<td>26</td>
<td>0.67</td>
<td>RSV 7 mL$<strong>; LSV 7.3 mL$</strong></td>
</tr>
<tr>
<td>Fujita\textsuperscript{16} (1994)</td>
<td>velocity mapping: (LV mitral inflow—Ao outflow)</td>
<td>TTE: jet area</td>
<td>29</td>
<td>RV 0.74</td>
<td>RV r=0.99$</td>
</tr>
<tr>
<td>Hundley\textsuperscript{46} (1995)</td>
<td>velocity mapping\textsuperscript††: (LV cine and aortic phase contrast)</td>
<td>catheterization: RF index</td>
<td>23</td>
<td>0.97</td>
<td>FSV 3±3$§</td>
</tr>
<tr>
<td>Kizilbash\textsuperscript{57} (1998)</td>
<td>velocity mapping\textsuperscript‡‡: (LV cine and aortic phase contrast)</td>
<td>TTE: RF pulsed Doppler RV</td>
<td>28</td>
<td>...</td>
<td>VM 0.6±4.8%</td>
</tr>
<tr>
<td>Kon\textsuperscript{58} (20050)</td>
<td>velocity mapping\textsuperscript††: (LV cine and aortic phase contrast)</td>
<td>RF gradient echo cine: RF with biventricular volumes</td>
<td>22</td>
<td>0.92</td>
<td>VM −2±7.7%§</td>
</tr>
</tbody>
</table>

n Indicates sample size; RF, regurgitant fraction; TTE, transthoracic echo; RSV, right ventricular stroke volume; LSV, left ventricular stroke volume; LV, left ventricular; Ao, aortic; RV, regurgitant volume; MR, magnetic resonance; FSV, forward stroke volume; TSV, total stroke volume; VM, velocity mapping; and GEC, gradient cine.

If not stated in the publication, statistics were calculated from the data provided in the manuscript. If this was not performed or data were not provided, a designation of "not available" (…) was given.

*Reported as mean difference±1 SD; if not available, the statistical test provided by the author is stated.

†RF (%)=LVSV−RSV/LVSV×100.
‡RF (%)=(TSV−FSV)/TSV×100.
§Interobserver reproducibility.
∥Intraobserver reproducibility.
\textsuperscript{††}RF (%)=LVSV−RSV/LVSV×100.
\textsuperscript{‡‡}RF (%)=LVSV−RSV/LVSV×100.
**Mean variation in RSV and LSV.
\textsuperscript{††}RVindex=TSVindex−FSVindex; RF=RVindex/TSVindex.
\textsuperscript{∥∥}RV=TSV−FSV; RF=RV/TSV.
pressure half-time valve area compared with echocardiography; however, a close correlation is seen between CMR and echocardiographic planimetered valve areas (Table 3). Planimetry of the mitral valve is a well-validated echocardiographic approach, because anatomy of the stenotic mitral valve results in a planar elliptical orifice in diastole. To find the minimal mitral or aortic orifice valve orifice on CMR images, thin (preferably 4 mm or less) overlapping slices are used as opposed to the usual 6- to 8-mm slices used for ventricular imaging. Reproducibility rates, as measured by interstudy, interobserver, and intraobserver variability of CMR for both aortic and mitral stenosis, are acceptable for clinical use (Tables 2 and 3).

**Evaluation of Valve Regurgitation**

CMR uses qualitative and quantitative indices of regurgitant severity similar to echocardiography: regurgitant jet area, regurgitant volume, and regurgitant fraction. Unlike stenosis severity, regurgitant severity in the past typically has only been clinically reported in qualitative or semiquantitative ways. With the endorsement of the 2006 American College of Cardiology/American Heart Association guidelines on valvular heart disease and the 2003 American Society of Echocardiography guidelines on valve regurgitation, clear recommendations are now available on how to quantify valve regurgitation.

**Regurgitant Jet Area**

The presence of a regurgitant jet can be visualized with cine pulse sequences, either SSFP or gradient echo. Flow turbulence across valves produces a loss of net signal (signal void) on cine pulse sequences due to dephasing of protons; however, correlation of the degree of signal void with the severity of regurgitation can be problematic, because the appearance of the signal void is highly dependent on pulse-sequence parameters. Early studies using CMR to evaluate valve regurgitation (aortic or mitral) focused on visualizing the signal void, with validations based on qualitative assessment or measurement of jet length or area compared with color Doppler or angiography (Tables 4 and 5). Evaluation of regurgitant severity based on jet area or length is no longer recommended, because this method is not a reliable indicator of disease severity. Nevertheless, visualization of distal flow disturbance by CMR, echocardiography, or angiography remains useful for detection of regurgitation and evaluation of jet direction and origin. Quantitation now focuses on proximal jet geometry, including vena contracta width and regurgitant orifice area.

**Regurgitant Volume and Fraction**

Regurgitant volume is defined as the amount of blood that flows in a retrograde direction across the valve with each heartbeat. Regurgitant fraction is the regurgitant volume expressed as a percentage of total stroke volume. CMR allows calculation of regurgitant volume by calculating the differences between right and left ventricular volumes, either by cine assessment of ventricular volumes or by velocity mapping and flow quantitation in the pulmonary artery and aorta. In the absence of cardiac disease, right and left ventricular stroke volumes are equal. With valve regurgitation, assuming only 1 valve is affected and no intracardiac shunt is present, the difference in stroke volumes reflects the regurgitant volume. Most adults with chronic valve regurgitation have significant regurgitation of only a single valve, which makes this approach widely applicable (Tables 4 and 5).

Phase contrast can directly quantify antegrade and retrograde flow volume across semilunar valves (Figures 4 and 5). To evaluate aortic regurgitation, phase-contrast imaging is performed in the aortic root, and total stroke volume and regurgitant volume are measured directly as the antegrade and retrograde transaortic flow volume rates. The same approach can be used for the pulmonic valve. Phase-contrast imaging of the mitral valve is more difficult because of the movement of the mitral valve annulus during ventricular systole. An alternative approach for measuring mitral regurgitation is to calculate total left ventricular stroke volume with SSFP imaging and forward stroke volume in the aorta using phase-contrast imaging. The difference between these 2 values represents the regurgitant volume. This technique is useful because in patients without mitral regurgitation, forward stroke volume has a very close correlation with total stroke volume.

Limited CMR studies on assessment of regurgitant orifice area (mainly by planimetry with aortic regurgitation) are available, even though this is 1 of the key variables in the assessment of regurgitant severity and the strongest predictor of clinical outcome. Planimetry of the aortic valve to determine the regurgitant orifice area has the same problem as does planimetry of the aortic valve for stenosis: The orifice does not lie in a single plane, and contraction of blood flow results in a smaller effective regurgitant orifice area than the anatomic area.

**Right-Sided Valve Disease**

**Pulmonary Valve Disease**

Pulmonic valve regurgitation is a major late complication after surgical repair of tetralogy of Fallot. Echocardiographic evaluation of pulmonic regurgitation severity is largely qualitative, based on vena contracta width and the density and deceleration slope of the continuous-wave Doppler signal. In addition, 2D imaging of the pulmonic valve in adults can be very challenging because of poor acoustic access. In the clinical setting, assessment of right ventricular function is only qualitative, because the complex shape of the right ventricle limits simple approaches to calculation of volumes and ejection fraction.

CMR provides visualization of the pulmonic valve and regurgitant jet, quantitation of regurgitant severity, and accurate measurement of right ventricular volumes and ejection fraction. Quantitation of pulmonic regurgitation by phase contrast has been validated against CMR right and left ventricular stroke volumes. Other studies are primarily limited to comparisons of quantitative CMR with qualitative echocardiographic measures, although 1 study showed a close correlation between regurgitant fraction by CMR and Dop-
ler pulmonary regurgitant severity defined by the diastolic duration of regurgitant flow on continuous-wave Doppler.62

CMR allows accurate quantitation of right ventricular volumes and ejection fraction because it allows direct visualization of the right ventricle in multiple tomographic planes and does not rely on geometric assumptions about chamber shape. In adults with chronic pulmonic regurgitation after tetralogy of Fallot repair, CMR has been used to measure the reduction in right ventricular volumes after valve replacement.63,64 One retrospective study suggested that right ventricular volumes may be an indicator of the optimal timing of pulmonic valve intervention.64 Prospective studies of this promising approach to the timing of pulmonic valve intervention are needed.

**Tricuspid Valve Disease**

Tricuspid regurgitation can be visualized on the basis of its signal void with cine imaging. Phase-contrast imaging of the tricuspid valve presents the same challenges as with the mitral valve owing to annular motion with the cardiac cycle. Tricuspid regurgitation can be calculated in terms of regurgitant volume and fraction in similar ways to mitral regurgitation: The forward stroke volume, as measured in the pulmonary artery with phase contrast, is subtracted from the total right ventricular stroke volumes from the SSFP images.

**Other Cardiac Imaging Approaches**

Computed tomography can provide accurate anatomic images of valves and valve motion. Computed tomography also provides precise quantitation of aortic valve calcification; however, quantitation of valve hemodynamics is predominantly limited to planimetry methods (valve area for stenosis and regurgitant orifice area for regurgitation).65–68

**Clinical Implications**

The primary advantage of CMR for assessment of valvular heart disease currently is its accurate and reproducible measurement of biventricular volumes and function in adults with chronic aortic, mitral, or pulmonic regurgitation. CMR may also provide a quantitative measure of valve stenosis and regurgitation, including regurgitant volume and fraction. As interventions for valve disease improve and are recommended earlier in the disease course, it becomes particularly important to ensure that patients do in fact have severe valve dysfunction and that our measures of ventricular size and function are precise and accurate. CMR is a promising approach for providing these data.

Despite the promise of CMR for evaluation of valve disease, some limitations still exist. Notwithstanding many published validation studies, these methods have not been widely used, and clinical experience is limited. CMR has not been shown to provide reliable information on pulmonary artery pressures or regurgitant orifice area, data that echocardiography does provide. In addition, echocardiographic criteria have been validated and compared with clinical outcomes, data that are not yet available for CMR measures of valve disease. Given the potential strengths of CMR, prospective studies using this approach to define disease progression and the optimal timing of intervention are needed. CMR may also provide a more accurate, precise, and reproducible measure of disease severity for trials of medical therapy in chronic valve disease.

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**References**


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