Magnetic Resonance Jet Velocity Mapping in Mitral and Aortic Valve Stenosis

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Background. Magnetic resonance (MR) phase-shift velocity mapping is an established method for measurement of nonturbulent intravascular flow. Shortening the echo time of the MR sequence to 3.6 msec allowed application of the technique to turbulent jet flow. The objective of this study was validation of MR jet velocity mapping in patients with cardiac valve stenosis.

Methods and Results. We used a 0.5-T Picker MR machine to measure peak poststenotic jet velocity in 15 consecutive patients recruited with known valve disease (six mitral stenosis, three of these restudied after valvoplasty, and 11 aortic stenosis). On the same day as the MR study, these patients underwent independent Doppler echocardiographic measurement of peak jet velocity. The results of 10 further MR investigations of aortic stenosis are also reported and compared with Doppler studies performed within 6 months. Of the 29 MR studies, 28 (97%) produced interpretable velocity maps, the one failure being attributed to misplacement of the imaging slice in a case of severe aortic stenosis. Agreement between MR and Doppler measurements of peak jet velocity in the recruited group was as follows: n=18; range, 1.4–6.1 m/sec; mean, 3 m/sec; mean of differences (MR–Doppler), 0.23 m/sec; standard deviation of differences, 0.49 m/sec.

Conclusions. In vivo MR peak jet velocity measurements agree well with those made by Doppler ultrasound. The technique, which is not subject to restricted windows of access and has potential for further refinements, could contribute to improved evaluation of stenoses, especially at locations where ultrasonic access is limited. (Circulation 1993;87:1239–1248)

Key words • imaging • valvular disease • ultrasound

The main aim of this study was in vivo validation of magnetic resonance (MR) phase-shift jet velocity mapping in patients with cardiac valve stenosis. We have previously published results of in vitro trials of the technique with preliminary clinical results.1,2 This method allows noninvasive mapping of the velocity distribution of turbulent jet flow either in or through a plane located anywhere in the heart and great vessels. Its main clinical role is probably in the assessment of obstructive lesions at positions where ultrasonic access is limited, for example, in surgically implanted ventriculopulmonary conduits.3 For clinical validation, however, we chose to recruit patients with mitral and aortic valve stenoses, with a wide range of jet velocities, in whom it was possible to obtain Doppler echocardiographic measurements of jet velocity with confidence of accuracy.4

We also report results of MR assessment of aortic stenosis in additional patients referred for MR imaging, since these help to illustrate the role of the technique as a supplement to echocardiography and/or invasive catheterization in aortic valve disease.

MR jet velocity mapping uses a gradient echo sequence with a very short echo time. Reduction of the echo time from 14 to 3.6 msec was the step that allowed application of MR phase-shift velocity mapping, already established as a robust method of in vivo flow measurement,5–11 to investigation of high-velocity jets.1,12,13 Gradient echo images acquired by use of sequences with longer echo times are subject to signal loss from turbulent regions (Figures 1A and 2B), and the extent of signal loss has itself been used as a semiquantitative method for assessment of stenosis, although this approach is unsuitable for accurate quantification.14,15 Shortening of the echo time minimizes signal loss from regions of turbulent flow. Recovery of MR signal from blood within the jet then means that phase-shift velocity mapping can be applied and, as in Doppler ultrasonic investigation and subject to the same provisos, the modified Bernoulli equation may be used to estimate the pressure gradient across the stenosis.16–19

Principles of Velocity Mapping

It is not essential to consider the theory of MR imaging in detail in this article, since other descriptions are available,20–23 but a brief explanation of certain principles underlying velocity mapping follows, with emphasis on the roles played by magnetic gradients.

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Figure 1. Above and facing page. Mitral stenosis (patient 3). Panel A: Diastolic frame from a pilot cine acquisition, echo time 14 msec, in vertical long-axis plane through mitral valve. The turbulent jet appears dark because of signal loss. Panel B: In-plane velocity map in the same plane but rotated to align the jet (dark) with the vertical read gradient direction. Panel C: Transjet velocity map, echo time 3.6 msec, velocity encoded through-plane in the direction of the slice-select gradient. The jet, of oval cross section, shows dark. A velocity profile is drawn across the jet, recording a peak velocity of 2.33 m/sec.
First, as in any type of MR imaging, magnetic gradients are used to encode positional information. The frequency of nuclear precession, which in this context means the frequency of resonance of the protons that constitute the nuclei of hydrogen, is proportional to the local magnetic field strength. So, when a gradient is applied, protons precess faster in stronger parts of the field than in weaker. Input of radio signal with a narrow band of frequencies will therefore energize the protons within a narrow slice of the field. This is the basis of slice selection. Subsequent gradients applied across and up and down the slice, known as phase encode and read gradients, are used to gain resolution in these directions.

In sequences used to measure flow, the magnetic gradients generally play two further roles, which depend on changes of precession frequency undergone by nuclei moving up or down an applied gradient.

Gradient switching can be used to elicit from energized nuclei the momentary peak of signal known as the echo. When a gradient is applied, the precession frequencies of nuclei located in stronger and weaker parts of the field diverge from one another. Reversal of the gradient reverses the effect, tending to bring nuclei back toward their original frequency and causing emission of a first “echo.” But at this moment, moving nuclei, because of changes of position within the gradients, fail to return to exactly their original frequency. For this reason, a further reversal of gradient is applied, and this results in a second echo, to which steadily moving as well as static nuclei contribute. It is during this second, or “even,” echo that the read gradient is applied, culminating the sequence. Signal is recovered from blood moving at a reasonably constant velocity, but accelerations and higher orders of motion, notably those within turbulent flow, can still lead to loss of signal caused by uncorrected frequency shifts at the instant of the even echo.

Finally, having acquired and localized signal from high-velocity blood, we come to the principle that enables velocity measurement. Moving nuclei, which are brought back at the end of the gradient switches to their original precession frequency, have, in the course of their movement through different field strengths, precessed faster or slower than protons in static tissue. These temporary frequency shifts can be used to bring about velocity-related phase shifts that can be measured accurately. For velocity mapping, a pair of phase maps is acquired, one for reference and the other velocity-encoded. Subtraction gives a velocity map in which phase variations caused by magnetic field inhomogeneities and susceptibility differences have been removed, leaving relatively pure velocity data derived from phase shifts. The direction of velocities mapped may be either in the image plane, encoded in the read gradient (Figures 1B and 2D) or phase encode gradient direction, or perpendicularly through the image plane, encoded in the direction of the slice-select gradient (Figures 1C and 2C).

The time from the peak of the initial energizing signal to the peak of the final echo is known as the echo time. A typical echo time for gradient echo sequences is 14 msec. Shortening the echo time reduces the period during which accelerating or turbulent flows bring about phase incoherence and loss of the final echo. Other factors also affect signal loss in positive and negative ways. For example, reducing the voxel size by increasing either the slice-select or read gradient strength will, on the one hand, reduce signal loss because the dephasing is over a smaller volume but, on the other hand, increase
signal loss because higher gradient strengths are required. The balance of these will depend on the type and distribution of flow. With complex flow, theory shows that by reducing the gradient durations, as well as by optimizing the waveform shape, a very significant reduction in signal loss is obtained. Our experimental results are in accord with this theory, demonstrating that as the echo time is reduced, the threshold of turbulence intensity at which signal is lost is very significantly raised. The shortest echo time we have been able to achieve, given the constraints of the available gradient coils and amplifiers, is 3.6 msec. This has proved to be sufficiently short to allow recovery of signal from all but the most intensely turbulent regions of flow, and jet core velocities of up to about 6 m/sec can be measured by use of this sequence.

<table>
<thead>
<tr>
<th>Patient</th>
<th>Sex</th>
<th>Age (years)</th>
<th>Mitral/aortic stenosis</th>
<th>Peak jet velocity (m/sec)</th>
</tr>
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<tr>
<td></td>
<td></td>
<td></td>
<td>MR1</td>
<td>MR2</td>
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<td>Recruited patients, MR and Doppler studies performed on same day</td>
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<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>1</td>
<td>F</td>
<td>67</td>
<td>MS, preballoon</td>
<td>2.0</td>
</tr>
<tr>
<td>2</td>
<td>M</td>
<td>18</td>
<td>MS, postballoon</td>
<td>1.7</td>
</tr>
<tr>
<td>3</td>
<td>F</td>
<td>46</td>
<td>MS, preballoon</td>
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<tr>
<td>4</td>
<td>F</td>
<td>50</td>
<td>MS</td>
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<tr>
<td>5</td>
<td>M</td>
<td>61</td>
<td>MS</td>
<td>1.9</td>
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<tr>
<td>6</td>
<td>F</td>
<td>51</td>
<td>MS</td>
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<td>7</td>
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<td>59</td>
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<td>11</td>
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<td>5.6</td>
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<td>12</td>
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<tr>
<td>14</td>
<td>F</td>
<td>73</td>
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<tr>
<td>Referred patients, MR and Doppler studies performed within 6 months</td>
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<td></td>
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<tr>
<td>15</td>
<td>F</td>
<td>35</td>
<td>AoS</td>
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<tr>
<td>16</td>
<td>M</td>
<td>74</td>
<td>AoS</td>
<td>4.9</td>
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<tr>
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<td>66</td>
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<td>4.2</td>
</tr>
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<td>M</td>
<td>31</td>
<td>AoS</td>
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<tr>
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<td>F</td>
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<td>AoS</td>
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<td>14</td>
<td>Homograft (+coarct)</td>
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<td>22</td>
<td>M</td>
<td>57</td>
<td>AoS (+coarct)</td>
<td>3.0</td>
</tr>
<tr>
<td>23</td>
<td>F</td>
<td>37</td>
<td>AoS (+coarct)</td>
<td>4.5</td>
</tr>
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<td>24</td>
<td>F</td>
<td>45</td>
<td>AoS (+coarct)</td>
<td>1.5</td>
</tr>
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</table>

MR, magnetic resonance; MR1, MR2, first and second observers’ measurements; MS, mitral stenosis; AoS, aortic stenosis; C-E valve, Carpentier-Edwards aortic valve prosthesis; +coarct, aortic coarctation also present.

**Methods**

**Subjects**

Twenty-nine separate MR studies of poststenotic jet velocity were made in a total of 25 patients 14–79 years old (mean age, 53 years). Fifteen of the patients (Table 1) were recruited prospectively for the study, being known to have mitral and/or aortic stenosis. Of these, five had mitral stenosis only, one had both mitral and aortic stenosis, and nine had aortic stenosis only. Three patients who had relatively severe mitral stenosis underwent balloon valvoplasty and were investigated both before and after the procedure. The remaining 10 subjects (Table 1) were patients referred for MR investigation in whom jet velocity measurement was used to assess aortic or subaortic stenosis. Doppler peak velocity measurements acquired within
6 months of the MR measurements were compared in eight of the patients.

To assess intraobserver variability in the interpretation of the MR jet velocity maps, all the MR jet studies were analyzed by a second observer familiar with the technique and unaware of the previous findings.

**MR Imaging**

We used a 0.5-T Picker International MR 2055 machine fitted with gradient coils with an internal bore of 530 mm; $G_{max}$, 14 mT/m; rise time, 400 μsec; and dB/dt$_{max}$, 8.75 T/sec, this maximal rate of change of field being within the British Radiation Protection Board's specified limit of 20 T/sec. A cine gradient echo sequence with full and partial velocity compensation was used for velocity mapping, with an echo time of 3.6 msec.

ECG gating was used to acquire 16 frames covering systole for aortic stenosis or diastole for mitral stenosis. Frame separation varied according to heart rate, with an average of 30 msec. Images representing slices 6–10 mm thick with a 40-cm field of view were acquired in 2×128 phase encoding steps.

Digital velocity information for each of the 128×128 points was represented visually by pixel intensity on a gray scale, black representing flow in the chosen direction and white in the opposite direction. The numerical value, i.e., the velocity, of each pixel on the image could be interrogated by movement of a cursor over the image or by plotting a curve or profile of velocity values along a selected line on the image.

A set of transaxial multislice spin echo images (Figure 2A) followed by oblique cine gradient echo images with 14-msec and 6-msec echo times (Figures 1A and 2B) were used for siting shots, working toward accurate identification of the location and direction of the poststenotic jet, deduced from the pattern of signal loss from the turbulent jet region. If doubt remained, a second cine acquisition, cutting orthogonally through the first, was used to locate the jet. A suitable plane for velocity mapping was then chosen. Each separate pilot acquisition took from 2 to 4 minutes, all the preliminary studies, with programming and processing, being completed in about 30–40 minutes.

For mapping of velocities in the jet plane, velocity was encoded in the read gradient direction, with the slice oriented so that this direction was aligned with the jet. Slices perpendicular to the jet were located immediately downstream of the orifice to avoid the region of high acceleration on the upstream side. For these transjet velocity maps, velocity was encoded in the slice-select gradient direction, i.e., through the image plane. We did not follow a strict sequence of acquisitions but rather proceeded according to the information available as the study progressed. If an in-plane velocity map was acquired first, this allowed accurate placement of the final transjet acquisition (Figure 1), or, alternatively, the transjet velocity map enabled accurate placement of the final in-plane acquisition (Figure 2). The second velocity map was acquired if time allowed or if the first proved adequate for location but not for measurement of the jet. It was advantageous to have a preliminary cine velocity map, showing both the shape of the jet and its movement through the course of systole, for optimum orientation of the final velocity map, located to measure the velocity of peak systolic flow. In cases in which there was a discrepancy between the peak velocities recorded in the two acquisitions, the higher value from the final, more accurately located acquisition was taken. Each velocity map took 6–9 minutes to acquire, depending on heart rate, bringing the total investigation time to between 45 and 80 minutes.

The velocity window, i.e., the range of velocities measurable without phase wrap, could be set to an appropriate value (e.g., 6 m/sec) by adjustment of the gradients used. When possible, a window about 50% greater than the predicted peak velocity was chosen so as to avoid aliasing but retain adequate sensitivity.

Once velocity maps had been obtained, movement of a cursor allowed interrogation of individual pixel values, or a “profile” of values along any chosen line could be drawn automatically (Figures 1C and 2D). Peak recorded jet velocity was estimated from an average of at least four pixels in the central jet core region.

**Doppler Echocardiography**

All patients had Doppler echocardiographic measurement of the peak poststenotic jet velocity performed on the same day as the MR study with a Hewlett-Packard 2-MHz continuous-wave Doppler probe used in conjunction with two-dimensional echocardiography and color flow mapping.

Each of these MR and Doppler studies was performed by an investigator experienced in the technique, without knowledge of previous velocity or pressure gradient measurements relating to the patient’s valve stenosis.

**Statistical Analysis**

MR and Doppler peak velocity measurements were compared (Figure 3A), and the differences were plotted against the means of each pair of measurements (Figure 3B) according to the method described by Bland and Altman. In these graphs and in their analysis, studies performed in recruited patients were distinguished from those performed on referred patients in whom MR and Doppler studies were performed on different days. The mean of each pair of measurements was taken as the best available standard, and the scatter of single measurements in relation to mean values was subjected to analysis. Three patients were excluded from analysis, one (patient 15) in whom MR peak velocity could not be measured because of failure to locate the jet and two (patients 18 and 20) for whom Doppler results could not be obtained, one for technical reasons and one because the patient was lost to follow-up. The MR peak velocity measurements by the two independent observers were analyzed similarly.

**Results**

Examples of MR jet velocity maps are shown in Figures 1 and 2. They allow measurement not only of peak velocities in the jet core but also of velocity distribution throughout the image plane and through the course of the cardiac cycle represented by the 16 frames of the cine acquisition.

Velocity maps in plane with the jet were generally easier to interpret because they clearly depicted the extent and location of the jet in relation to upstream...
and downstream structures. In velocity maps transecting the jet, the jet core, passing through the plane, was represented by only a small region composed of few pixels. This provided little visual information on the anatomic location of the jet, but the shape of the transected jet core was taken to indicate the shape of the valve orifice. This was ovoid in the two cases illustrated or flattened in cases with a more slitlike orifice.

There was variation in quality of the velocity maps. On the basis of previous in vitro observations, we attributed this to variable accuracy of slice location in relation to the jet core. Well-placed, in-plane velocity maps show the length of the jet clearly, with acceleration and deceleration zones upstream and downstream. A slightly misplaced velocity map gives a more fragmented, discontinuous picture.

Of 29 MR studies, only one failed to provide a jet velocity measurement because of failure to locate the jet. Six more (in patients 4, 6, 10, 11, 12, and 14) provided velocity maps of suboptimal quality, but peak velocity measurements from these MR studies were included in the analysis.

The comparison of peak jet velocities measured by MR and Doppler is presented in Figure 3 and in Table 2.

In three of the nine nonrecruited patients (patients 19, 22, and 24), velocity mapping drew attention to subvalvar or valvar stenosis of a severity that had not previously been documented for the patient but that was subsequently confirmed by Doppler or catheter investigation. In one case (patient 16), there was doubt about the accuracy of Doppler assessment of the jet passing obliquely through the distorted and calcified valve, and attempted catheterization had failed to cross the valve, so jet velocity mapping was felt to be the most reliable.

**Table 2. Comparison Between Magnetic Resonance and Doppler Assessment of Peak Jet Velocities**

<table>
<thead>
<tr>
<th>Studies</th>
<th>n</th>
<th>Range (m/sec)</th>
<th>Mean (m/sec)</th>
<th>Mean of differences</th>
<th>SD of differences</th>
<th>% SD</th>
</tr>
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<tbody>
<tr>
<td>All</td>
<td>26</td>
<td>1.2–5.5</td>
<td>3.1</td>
<td>−0.10</td>
<td>0.46</td>
<td>15.5</td>
</tr>
<tr>
<td>Same day</td>
<td>18</td>
<td>1.4–6.1</td>
<td>3.0</td>
<td>−0.23</td>
<td>0.49</td>
<td>15.6</td>
</tr>
<tr>
<td>Within 6 months</td>
<td>8</td>
<td>1.2–5.0</td>
<td>3.6</td>
<td>0.18</td>
<td>0.23</td>
<td>9.1</td>
</tr>
</tbody>
</table>

**Figure 3.** Panel A: Peak jet velocity measurements by magnetic resonance (MR) plotted against those by Doppler ultrasound. Panel B: Difference between MR and Doppler peak velocity measurements plotted against the mean of each pair. See text for statistical analysis. MS, mitral stenosis; AoS, aortic stenosis.
Discussion

Agreement Between Techniques

The agreement between measurements of peak jet velocity by MR velocity mapping and by Doppler ultrasonography, in both the recruited and the referred patients, indicates that MR has sufficient accuracy to be of clinical value. The distribution of points in Figure 3B suggests that agreement is slightly better for jets with velocities below 5 m/sec and that agreement is not as good above 5 m/sec, where the jet is likely to be narrow and intensely turbulent. Potential sources of inaccuracy have been described elsewhere. The one failure to obtain a usable velocity map was in a patient in whom Doppler ultrasonography had recorded a peak velocity of 6.2 m/sec, and this failure was probably attributable to misplacement of the slice in relation to the narrow jet. In this patient, the unusual degree of signal loss seen on the short echo time cine magnitude (signal intensity) images gave a qualitative indication of the severity of stenosis. More accurate placement of the velocity mapping plane would probably have enabled jet velocity measurement, and this underlines the dependence of the present technique on absence of patient movement and meticulous imaging technique.

There is slightly better agreement in studies on the eight nonrecruited patients, in whom MR and Doppler investigations were performed on different days. The patients in this group were all investigated after studies in the recruited group had been completed, and the improved agreement correlates with improved quality of MR velocity maps, probably because of improved accuracy of slice location.

Agreement between the two observers interpreting the same MR velocity maps was, as expected, closer than that between MR and Doppler results. Repetability of the MR velocity mapping procedure was not assessed, nor did we compare the relative advantages of in-plane and through-plane jet velocity mapping. In several cases in which both were acquired, especially during the earlier part of the study, the first velocity map was imperfectly located and served only as an additional sitting shot for the final acquisition.

We did not study the accuracy of Doppler echocardiography, which is a well-established clinical technique with a high level of accuracy and reproducibility when used for measurement of peak jet velocity in mitral and aortic valve stenosis.4

Clinical Importance

The importance of this study is its clinical validation of a technique that is likely to have its main role in other groups of patients. Ultrasonic access in adults with congenital heart disease, many of whom have undergone surgery at least once, can be limited. This includes patients with obstructing ventriculopulmonary conduits inserted for correction of Fallot's tetralogy or pulmonary atresia, patients with coarctation or recoarctation of the aorta, patients with intra-atrial baffle obstruction after a Mustard operation, and patients with obstruction after a Fontan procedure. The transesophageal approach, especially with biplane or multiplane transducers, improves the reach and quality of echocardiographic studies, but extracardiac obstruction may still be difficult to assess. Doppler alignment with a jet depends on the location of the transducer, whereas MR velocity measurement can be aligned with jets of any orientation.

In stenoses of native valves, it is usually possible to obtain diagnostically useful jet velocity information using Doppler ultrasound, although there are occasional cases, as illustrated by patient 16 in this study, in which calcification and deformation of aortic valve cusps not only prevents adequate Doppler assessment but also can result in failure to cross the valve at catheterization. In such cases, MR jet velocity mapping provides an alternative. In addition, it provides information on jet size and shape and on the distributions of velocities in and around the jet more comprehensively than ultrasound.

Limitations of MR Velocity Mapping

The limitations of cine MR velocity mapping include the relatively long acquisition times; the poor quality of cardiac gated images in patients who have cardiac arrhythmias; the confined bore of the magnet, which may preclude use of the technique in patients with claustrophobia; and the high cost of the machine. These limitations may become less important with the development of real-time MR velocity mapping using an echo planar or other rapid imaging techniques,28-30 open access magnets, and dedicated cardiovascular machines.

Prosthetic Valves and MR Imaging

Prosthetic valves of many different designs are now in widespread use. Among the eight referred cases, one patient had an aortic homograft and one a Carpentier- Edwards aortic xenograft. Evaluation of prosthetic valves by catheter can be more difficult than with native valves, and morbidity is greater. The prosthetic material itself, with a low proton density, gives little MR signal, and, if metal is incorporated, distortion of the magnetic field leads to localized image defect. The defect is small for spin echo images but larger in gradient echo images, making it difficult to assess jet velocities in the immediate locality of the valve. Metal valves are not ferromagnetic, however, and at currently available magnetic field strengths there is no danger of the field's exerting significant mechanical forces on the valve.31-34

Future Potential of MR Velocity Mapping

The potential of MR velocity mapping for further development is well worth considering. MR has unrivaled capacities for acquisition of multidimensional and multi-

TABLE 3. Variability of Interpretation of Magnetic Resonance Peak Jet Velocity by Two Observers

<table>
<thead>
<tr>
<th>n</th>
<th>Range (m/sec)</th>
<th>Mean (m/sec)</th>
<th>Mean of differences</th>
<th>SD of differences</th>
<th>% SD</th>
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<td>28</td>
<td>1.5-5.6</td>
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<td>0.11</td>
<td>0.29</td>
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directional velocity data.35–37 Velocities in a pulsatile flow field are distributed in the four dimensions of space and time, and measurement at each point of three orthogonal components is necessary to determine the local velocity vector. In this study, to economize on acquisition time and to limit the quantity of data to be handled, velocity measurements were confined to one direction, aligned with the jet, distributed through the two dimensions of a plane and through the course of the cardiac cycle. Choices had to be made to fit in with the limitations of the available hardware and software. Current developments in rapid imaging,38 together with improvements in data storage and handling, mean that acquisition of comprehensive (four-dimensional, three-directional) flow data for a volume containing a stenosed or, for that matter, a regurgitant valve may be feasible. This should enable calculations of the volume of flow approaching an orifice to be correlated with peak jet velocity, allowing calculation of orifice area. It should also allow analysis of jet shape and downstream flow, which could have a bearing on the applicability or lack thereof of the modified Bernoulli equation and might lead to development of new formulas for calculating the rate of energy dissipation for a given rate of flow through a particular stenosis, all of which could be relevant to decision making on the timing and type of intervention.

References


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