Calibration of color Doppler flow mapping during extreme hemodynamic conditions in vitro: a foundation for a reliable quantitative grading system for aortic incompetence

DONALD F. SWITZER, M.D., AJIT P. Yoganathan, Ph.D., NAVIN C. NANDA, M.D., Y-R WOO, Ph.D., AND ANDREA J. RIDGWAY, B.S.

ABSTRACT If color Doppler imaging is to continue to evolve into a reliable clinical method to noninvasively evaluate regurgitant lesions, then its grading methods must be quantitated and calibrated under extreme hemodynamic conditions. A left heart pulse duplicator was used to provide a completely controllable system to study aortic incompetence jet morphologies as a function of hemodynamic extremes. The system was first used to calibrate the limits of resolution of color Doppler imaging. Next, to define which jet features reliably predict the defect size or the regurgitant fraction and which are primarily influenced by instantaneous hemodynamic variables, we measured the jets’ maximal length, width, proximal width, and temporal pattern of color variance during independent variations in the heart rate, cardiac output, and pressure gradient across the incompetent valve. The proximal jet width (immediately below the valve plane) was the only reliable independent predictor of both the defect size and the regurgitant fraction. Jet depth accurately predicted peak velocity (quantitated by laser Doppler velocimetry); it reliably predicted the severity of incompetence only at a known pressure gradient across the valve. Large defects (5 mm) produced jets with maximal color variance in early diastole, whereas small defects produced pandiastolic variance. *Circulation*, 75, No. 4, 837–846, 1987.

QUANTIFICATION of native and prosthetic valvular incompetence has traditionally relied upon contrast angiography. Besides the requirement for an invasive catheterization, several limitations make this an imperfect gold standard and it is impractical for serial examinations. Conventional pulsed Doppler flow mapping has proven to be a valuable clinical alternative to facilitate the noninvasive detection and semiquantification of regurgitant flow. From the Division of Cardiovascular Disease, University of Alabama at Birmingham, and the School of Chemical Engineering, Georgia Institute of Technology, Atlanta.

Address for correspondence: Dr. Navin C. Nanda, University of Alabama at Birmingham, Heart Station, SWB/W001, Birmingham, AL 35294.

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Doppler study can reliably substitute for an invasive catheterization.

However, it is not known how extreme hemodynamic conditions will influence the morphology of regurgitant jets, as seen with color Doppler imaging. It is to be expected that instantaneous variations in hemodynamic variables will alter the jet’s appearance and therefore significantly affect the grading of regurgitation between serial examinations if the influence of these factors is not appreciated by the examiner. Furthermore, with the convention of grading maximum jet dimensions (rather than mean), this is particularly relevant for beat-to-beat variability induced by dynamic physiologic variables. This has likely contributed to the considerable variance in the reported sensitivity and specificity for detection and quantification of regurgitant lesions by color Doppler imaging. Before color flow mapping can approach its full clinical potential and seriously compete with angiographic grading of regurgitant lesions, characteristics of regurgitant
jet morphology must be "calibrated" in a controlled system under extreme hemodynamic conditions.

It is remarkable that there are as yet no universally accepted criteria for the grading of regurgitant lesions by either conventional pulsed or color flow mapping. Most laboratories grade aortic incompetence (AI) primarily on the basis of the maximal length of the regurgitant jet. Other characteristics that have been given varying amounts of attention include the jet's maximum width, area, color variance (turbulence), diastolic duration, and the ratio of the jet's area (or length) to that of the left ventricle.

The purpose of this study was to use a completely controllable and reproducible system in vitro to determine which are the most reliable characteristics of an AI jet to use for grading of regurgitation by color flow mapping. We used a left heart pulse duplicator containing incompetent prosthetic valves to first calibrate limits of resolution of color Doppler imaging. Hemodynamic variables were then independently varied to differentiate the aspects of a regurgitant jet's morphology that reliable predict the severity of incompetence from features that primarily reflect the hemodynamic profile imposed across the incompetent valve. It was hoped that this information could then be applied clinically to serve as the foundation for a reliable, standardized grading system for noninvasively quantitating valvular incompetence.

**Methods**

**Pulse duplicator.** The left heart pulse duplicator system was designed to simulate physiologic pressure and flow waveforms in vivo. Complete details on the flow apparatus have been reported previously and are illustrated in figure 1. 15–19 Prosthetic valves were mounted within the aortic flow chamber (figure 2). The system was then filled, at room temperature, with a water glycerine solution to simulate the viscosity (3.5 cp) and density (1.05 × 10³ kg/m³) of blood. A corn starch suspension (approximately 1% by weight) containing 10 μm particles was added to provide adequate scattering of the Doppler signals and thereby simulate the ultrasonic reflectivity of red blood cells. This pulse duplicator system allows for independent control over preload, afterload, heart rate, and cardiac output. For each incompetent valve model studied, the 48 sets of hemodynamic variables outlined in table 1 were used. Preload, aortic compliance, and systemic resistance were then adjusted to maintain each selected cardiac output and mean aortic pressure (MAP). Ventricular compliance was adjusted to maintain the left ventricular end-diastolic pressure (LVEDP) within the physiologic range (<15 mm Hg) whenever possible. As expected, with severely incompetent valves (large defects subjected to an elevated pressure gradient) it was not possible to maintain either a normal LVEDP or an aortic end-diastolic pressure (AoEDP) greater than 80 mm Hg, regardless of the selected ventricular compliance and systemic afterload.

**Prosthetic valves.** Hancock and Ionescu-Shiley bioprosthet-
ic valves were first imaged by color Doppler de novo in the aortic flow chamber under the above extreme hemodynamic conditions. Defects of 1, 3, or 5 mm diameter were then punched into one leaflet of each valve and imaging was repeated for each incompetent valve during the same 48 combinations of hemodynamic conditions.

Starr-Edwards, Björk-Shiley, and St. Jude’s mechanical valves were also mounted in the aortic flow chamber and their flow patterns were examined during extremes of MAP (50 to 200 mm Hg) and cardiac output (1.5 to 6.5 liters/min) to observe the dependence of their “physiologic” incompetence (related to normal closure of the mechanical leaflets) on variations in hemodynamic variables. “Pathologic” incompetence of these valves was then induced by obstructing closure of a leaflet with a mock thrombus. The valves were then reexamined by color Doppler imaging during the same hemodynamic extremes.

**Color Doppler flow mapping.** Color flow mapping of AI was performed with the Irex-Aloka 880 instrument. As shown in figure 2, regurgitant jets were imaged with a 2.5 MHz transducer placed on the left ventricular chamber (to simulate an apical window) 8 cm away from the plane of the prosthetic aortic valve. A 9 or 12 cm sample depth was used. Systolic and diastolic timing was provided on the Doppler images by use of a simultaneous flow curve from the aortic flow probe. Color gains were set to a level immediately below that which resulted in minimal background color “noise.” To provide reliable jet dimensions, considerable emphasis was placed on optimizing the transducer’s angulation to provide a longitudinal transection of each jet’s full profile from the plane of the valve to the distal tapering tip of the jet.

The regurgitant jets were measured with the instrument’s online calipers for maximal depth, maximal width, and minimal proximal width (minimal diameter within the first centimeter of the jet, immediately below the valve plane). All reported measurements are the mean of 5 beats (during each set of conditions). A qualitative assessment was also made of the jet’s relative amount of turbulence in early vs late diastole. These observations were then repeated during each of the 48 sets of hemodynamic conditions outlined in table 1 for each incompetent mechanical or bioprosthetic valve (with 1, 3, or 5 mm defects).

**Lasers Doppler velocimetry.** The dimensions obtained with color flow mapping were compared with those obtained with the very precise technique of laser Doppler velocimetry performed on the same jets (under identical hemodynamic conditions). The laser Doppler used (DISA 55x modular system) provides a two-dimensional velocity profile with a spatial resolution of 0.3 mm, and thus provides a very precise profile of the jet’s dimensions and maximal velocity. A detailed description of the laser Doppler flow system has been described previously. 15-17

**Regurgitant fraction.** The regurgitant fraction (RF) was calculated from the forward and reverse flow volumes measured by a Carolina Medical 25 mm cannulating type electromagnetic flow probe that was interfaced to a FM501 electromagnetic flowmeter. The analog signal outputs from the flowmeter were fed into an Apple II Plus microcomputer via a 16-channel analog-to-digital converter. The analog signal was digitized at the rate of 500 to 1000 samples per second and analyzed on-line (i.e., beat-to-beat) by the microcomputer. The microcomputer

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**TABLE 1**

<table>
<thead>
<tr>
<th>Hemodynamic variables</th>
<th>Varied</th>
<th>Constant</th>
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<tbody>
<tr>
<td>MAP/EDP (mm Hg)</td>
<td>HR (bpm)</td>
<td>CO (l/min)</td>
</tr>
<tr>
<td>50/45</td>
<td>50</td>
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<td>200/130</td>
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</table>

MAP, heart rate (HR), and cardiac output (CO) were each independently varied, keeping the other two variables constant (as shown). Then the CO was changed for each MAP (with HR remaining at 70), producing a total of 48 sets of conditions for each valve. EDP = aortic end-diastolic pressure.

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**FIGURE 2.** Detailed view of the simulated aortic flow section. The Doppler transducer is placed 80 mm from the aortic valve (AV) plane, simulating a left ventricular (LV) apical view of the AI jets. The pressure taps monitor instantaneous transvalvular gradients, and the aortic flow probe is used for deriving the RF. All dimensions shown are in millimeters.
was programmed to calculate the net forward and net reverse flow volumes per beat, with an accuracy of $\pm 1$ ml per beat.\textsuperscript{20}

**Resolution.** The imaging limits of resolution of the color Doppler instrument were also evaluated. To determine the minimal amount of net flow that it could detect (axial flow resolution), cardiac output was decreased in 0.25 liter/min decrements until no flow through a normal prosthetic valve could be detected. To determine the limits of lateral resolution (for differentiating adjacent jets), color imaging was first performed on a bioprosthesis having two 3 mm defects punched in two separate leaflets. A second valve was then imaged that had two 3 mm defects punched only 3 mm apart on the same leaflet to determine if these jets could also be differentiated.

**Color variance.** Finally, to evaluate color flow mapping's reliability in predicting disturbed flow on the basis of color variance, streak photography was performed during imaging of jets from 3 and 5 mm defects in bioprothetic valves. Streak photography provides a qualitative flow map and can be used to characterize a jet's flow pattern as laminar or turbulent. These patterns were then compared to those predicted by the color profile provided by Doppler imaging (under the same hemodynamic conditions).

**Statistical analysis.** The data were subjected to statistical analysis with the use of the Pearson product moment correlation coefficients and the estimated slopes and intercepts. Multiple linear regression was performed for models having two or more independent variables. The predictions of regurgitant fraction and defect size were tested with linear discriminant analysis. Finally, Spearman correlation coefficients were determined for the same conditions. There was no significant difference between the Pearson and Spearman coefficients, and therefore the former are reported in the data.

**Results**

**Resolution.** Net forward flow could be detected (in this system) with a cardiac output as low as 0.5 liter/min. Aortic incompetence could be detected without difficulty through the 5 mm and 3 mm defects, as well as through the 1 mm pinhole, provided that the cardiac output exceeded 1.5 liters/min and the MAP was at least 75 mm Hg. As will be shown subsequently, the dimensions of the AI jet, and therefore color Doppler's sensitivity for detecting AI, were most dependent on the pressure gradient ($\Delta P$) across the valve. This was particularly relevant for the 1 mm pinhole. The sensitivity of color imaging for the detection of AI was compromised by tachycardia (heart rate $\geq 110$ beats/min) or a low cardiac output ($\leq 3.5$ liters/min).

During the evaluation for lateral resolution, the two jets originating from defects placed in separate leaflets were always easily differentiated. Their jets were separated laterally by up to 20 mm on the color Doppler image. Furthermore, even the two jets originating from two defects placed only 3 mm apart on one leaflet could be differentiated (figure 3, A), regardless of the hemodynamic settings. However, when subjected to a significantly elevated MAP ($>150$ mm Hg), the maximum width of the two jets increased and obscured the fact that these were two distinct jets.

**Hemodynamic effects on jet morphology.** Analysis of AI jet morphology (from any size defect) was optimal at slower heart rates. This was particularly relevant when imaging jets from small defects (1 mm) during a low MAP ($<100$ mm Hg) and/or a cardiac output of 3.5 liters/min or less. To optimize jet analysis, we therefore performed most of the imaging with the heart rate fixed at a physiologic rate of 70 beats/min (table 1). Jet dimensions, velocity, and RF are shown as a function of $\Delta P$ and defect size in table 2. With the gain settings and transducer orientation described, there was negligible variation ($\leq 1$ mm) in the five measurements obtained for each proximal or maximal width.

**Jet depth.** The reliability of the maximal jet depth (or length) as an independent variable for predicting either the amount of regurgitation or the size of the valve's incompetent orifice (defect size) was determined. As expected, this depth was clearly greater for jets arising from a larger defect ($5 > 3 > 1$ mm; figure 4). However, it also demonstrated that the jet depth has a very strong dependence on the $\Delta P$ ($p = .0001$ to .012). Specifically, a jet of a given length (25 to 40 mm) could originate from either a large defect during a low or normal MAP, or from a small defect if subjected to a large $\Delta P$ (AoEDP $\geq 95$ mm Hg). The RF also could not be reliably predicted by use of jet depth. The RF could be predicted from jet depth with statistical confidence only at a known defect size (figure 4). Jet length did predict peak flow velocity. Figure 5 demonstrates the linear relationship between the AI jet velocity measured by laser Doppler velocimetry (at 12 mm from the valve plane) and the jets' maximum depth measured by color Doppler imaging ($p = .0001$) under identical hemodynamic conditions.

**Maximal width.** The jets' maximal width was not an independent predictor of either the defect size or the RF. Figure 6 demonstrates the poor correlation found between the maximum jet width and the RF ($p = .1099$). Although the maximum width was greater for jets arising from larger defects, there was a stronger dependence of width on the imposed $\Delta P$ ($p = .0003$ to .029). Hence, wider jets (>10 mm) occurred with either a large defect, or with small defects subjected to an increased pressure gradient (AoEDP $\geq 95$ mm Hg).

**Proximal width.** The jets' minimal proximal width (measured within the first centimeter immediately below the valve plane) was predictive of the size of the defect (figure 7). This dimension was 100% specific in differentiating the three defect sizes, provided that the AoEDP was greater than 80 mm Hg. With a lower AoEDP, the proximal width demonstrated a significant dependence on the aortic pressure and did not
reliably predict the defect size. However, there was a significant relationship (p = .0004) between this proximal width and the RF that was independent of the ΔP at all aortic pressures (figure 7). For this measurement to be reliable, it was crucial that the color flow imaging plane be aligned through the jet’s long axis and include its origin from the valve (figures 3, B to D), since misleading morphologies resulted if imaging planes that transected the jet several mm below the valve were considered.

**Color variance.** The temporal profile of color variance (signifying disturbed flow) often provided additional information useful in grading the severity of AI, particularly in the presence of an elevated ΔP. With small defects, such color variance persisted throughout the jets’ duration, whereas with the 5 mm defect, color variance was limited to very early diastole; the jet then became a homogeneous red or orange color for the remainder of diastole (figure 3, C and D). Streak photography was used to validate the use of these color changes in predicting flow patterns. Qualitative comparisons confirmed that when the color image predicted flow variance, streak photographs demonstrated significantly disturbed flow, whereas monochromatic homogeneous color jets correctly predicted undisturbed laminar flow. Also, both the laser and continuous-wave Doppler examinations showed that the AI emerging from large defects had their greatest velocity in early diastole, whereas the velocity of jets from small defects did not significantly decrease throughout diastole.

**Mechanical valves.** With the heart rate set at 70 beats/...
TABLE 2
Results

<table>
<thead>
<tr>
<th></th>
<th>MAP (mm Hg)</th>
<th>p value (vs MAP)</th>
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<tr>
<td></td>
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<tr>
<td>3 mm defect</td>
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<tr>
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<td>Laser velocity</td>
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<td>5 mm defect</td>
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Jet dimensions (in mm), RF, and the laser Doppler-derived velocities (cm/sec) are shown for AI jets arising from 1, 3, and 5 mm diameter defects as a function of the MAP. The RF from the 1 mm defect was not large enough to be reliably calculated. The p values show the relationship between each variable and the MAP. For this series, the cardiac output was fixed at 4.5 l/min and the heart rate was 70 bpm.

*For the proximal jet width (proximal width) p value determinations, widths at a MAP of 50 and 75 mm Hg were excluded (see text).

Defect of a given size? If color flow mapping is to continue to evolve into a clinically valuable and reliable tool to detect, quantitate, and follow the course of regurgitant lesions, then the impact of instantaneous hemodynamic variables on the appearance of regurgitant jets must be defined and quantitated in a controlled, reproducible system. This information may then be applied to derive a reliable grading system for AI.

The instantaneous hemodynamic state has a profound influence on most of the jet’s dimensions. The maximum jet depth is the primary measurement used

min and cardiac output at 4.5 liters/min, the “physiologic” closure-related AI from mechanical valves was most pronounced when the MAP was 125 mm Hg or less; it became undetectable at a greater ΔP. With the MAP fixed at 100 mm Hg and heart rate at 70 beats/min, the depth of the jet showed an inverse relationship to changes in the cardiac output (table 3). Furthermore, this physiologic AI was always limited to early diastole and its maximal jet length was always under 2 cm, regardless of loading conditions. This was always clearly distinguishable from pathologic incompetence (induced by placement of a mock thrombus at a leaflet), which was always pandyastolic, had a direct relationship to both MAP and cardiac output, and generally extended greater than 2 cm (particularly with a MAP ≥125 mm Hg).

Discussion

Which aspects of a regurgitant jet’s morphology can be used by the color Doppler method to reliably predict the size of the defect in an incompetent aortic valve without being significantly influenced by variations in hemodynamic variables? Also, which features will predict the severity of regurgitation (volume) through a

FIGURE 4. A, Although the regurgitant jet depth is proportional to the size of the defect in the aortic valve leaflet (5 > 3 > 1 mm diameter defects), there is also a very strong dependence (p = .0001 to .012) of this depth on the imposed instantaneous ΔP. The shaded area emphasizes that a jet of 20 to 40 mm in depth could originate from a 1, 3, or 5 mm defect, depending on the ΔP. B, The RF also could not be predicted independently from the jet depth. The statistically significant relationships shown between jet depth and RF exist only when the depths are interpreted at a known defect size (3 and 5 mm examples shown). Given a jet of a specific depth, the RF can only be predicted if the defect size is known. This would therefore not be generally applicable clinically. EDP = aortic end-diastolic pressure.
by most laboratories for both color and pulsed Doppler grading of AI. We have shown by laser Doppler velocimetry that this length has a linear relationship to the jets' peak velocity (figure 5). Furthermore, this velocity (and therefore jet depth) has a strong dependence on the pressure gradient (p = .0001 to .012). It is not independently predictive (p = .0604) of the size of the valve's defect (figure 4). At any given ΔP, the jet from a larger defect will extend deeper into the left ventricle. However, even the 1 mm pinhole emitted a jet that extended up to 4 cm when subjected to an elevated MAP.

Furthermore, the severity of aortic incompetence (the RF) does not necessarily correlate with the jet length. The RF depends on the size of the defect, the left ventricular compliance, the pressure gradient across the defect, and the duration of this gradient (a function of both the heart rate and the rate of rise of the left ventricular diastolic pressure). Using this controlled system in vitro, we have found that, to be useful, the jet length must be interpreted at a known ΔP and an approximated defect size. For example, with acute AI, a small defect subjected to an elevated ΔP will emit a jet that has a high velocity primarily limited to early diastole. This would not produce a large volume of AI but is often associated with a deep jet (>2 cm) immediately after valve closure and a considerably shorter jet through the remainder of diastole (depending on the ventricular compliance). Furthermore, if the regurgitant volume is out of proportion to the RF

(As in high-output states involving chronic AI into a large, compliant left ventricle) jet length will overestimate the RF. Therefore, the maximal jet length should not be used as an independent criteria for grading the severity of AI in any Doppler study.

In a similar manner, the maximal downstream jet width was found to have a strong dependence on both the defect size and the ΔP (figure 6). Therefore, it may also be used to facilitate grading of AI, but only at a known ΔP. As an independent variable, downstream width does not predict the defect size (p = .4865) or regurgitant volume (p = .1099). Jet area is a third

**FIGURE 5.** A precise peak velocity measurement was obtained by laser Doppler velocimetry and this was compared with the color Doppler-derived jet depth. A significant relationship was demonstrated between jet depth and peak velocity, and unlike that for RF, this relationship remained significant even when the defect size was not given (p = .0001).

**FIGURE 6.** A. There was a significant relationship between the maximum jet width (max width) and the ΔP across the valve. However, this width could not be used to differentiate between the three defect sizes (1, 3, and 5 mm diameter) because of its greater dependence on the ΔP and because of the overlap shown between the defect sizes as the ΔP was varied. B. There was also no association (p = .1099) between the maximal jet width and the measured RF.
The minimal proximal jet width (prox width) measured immediately below the valve plane was 100% specific in differentiating the three defect sizes (1, 3, and 5 mm diameter), provided that the MAP was at least 100 mm Hg (shaded areas). This dimension was unique in that it showed minimal variation as a function of increasing the ΔP, provided that the aortic end-diastolic pressure (EDP) was 80 mm Hg or more. With a decreased EDP, this dimension had significant dependence on the aortic pressure. B. The minimal proximal width (prox width) was, however, highly predictive of the RF, independent of the aortic pressure. Plotted are the proximal widths from the same jets shown in A. The jet with a 2 mm width occurred during an EDP of 45 mm Hg. The dependence of the proximal width on the RF is therefore reliable, even with a low aortic EDP (as seen in patients with chronic AI). This dimension is therefore the most reliable predictor of both the defect size and the RF.

Variable sometimes quantitated for grading AI. Since it is determined by the jet’s maximal length and width, it will also be strongly dependent on both the instantaneous hemodynamic state and the defect size.

There were features that did predict the RF and the presence of large defects independently of the imposed hemodynamics. The most reliable predictor was the minimal width of the proximal jet (adjacent to the valve). As illustrated in figure 8, the proximal first centimeter of jets from small defects tapers to a width significantly less than their maximal downstream width (as in figure 3, B), whereas jets from larger defects have a more cylindrical proximal shape (figure 3, C and D). This distinction was true regardless of extremes in the cardiac output and MAP (figure 7). Although this jet width was 100% specific in different-

**FIGURE 7.** A, The minimal proximal jet width (prox width) measured immediately below the valve plane was 100% specific in differentiating the three defect sizes (1, 3, and 5 mm diameter), provided that the MAP was at least 100 mm Hg (shaded areas). This dimension was unique in that it showed minimal variation as a function of increasing the ΔP, provided that the aortic end-diastolic pressure (EDP) was 80 mm Hg or more. With a decreased EDP, this dimension had significant dependence on the aortic pressure. B, The minimal proximal width (prox width) was, however, highly predictive of the RF, independent of the aortic pressure. Plotted are the proximal widths from the same jets shown in A. The jet with a 2 mm width occurred during an EDP of 45 mm Hg. The dependence of the proximal width on the RF is therefore reliable, even with a low aortic EDP (as seen in patients with chronic AI). This dimension is therefore the most reliable predictor of both the defect size and the RF.

**FIGURE 8.** Schema showing the overall dependence of regurgitant jet morphology on the defect size and the hemodynamic state. The maximum jet width and length are strongly dependent on both the ΔP and the defect size. The RF can be predicted from these dimensions if the approximate ΔP is known. However, the minimal proximal width is not significantly influenced by the ΔP, and is therefore the only reliable independent predictor of both the size of the valvular defect and the RF. B, The basic morphology observed for jets arising from each defect size.

**TABLE 3**

<table>
<thead>
<tr>
<th>Mechanical valves</th>
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<tbody>
<tr>
<td>Fixed CO at 4.5 l/min</td>
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<tr>
<td>MAP</td>
</tr>
<tr>
<td>Depth</td>
</tr>
<tr>
<td>Fixed MAP at 100 mm Hg</td>
</tr>
<tr>
<td>CO</td>
</tr>
<tr>
<td>Depth</td>
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</table>

The depths of jets (in mm) from a normal closing Starr-Edwards valve are shown as a function of MAP and cardiac output (CO). Heart rate was kept fixed at 70 bpm.
iating the three defect sizes, it was consistently wider than the actual diameter of the anatomic defect in the valve; therefore, it should be used to predict the approximate, not the exact defect size. This probably relates to the inability to image the jet before it begins to widen, and to the formation of small flow eddies around the perimeter of the high-velocity AI jets as they emerge from the defect and "spray" into the low pressure, early-diastolic ventricle. This would be particularly relevant in the setting of generous color gain settings.

The jet's proximal width was also a strong predictor (p = .0004) of the RF (figure 7). This is further supported by an independent clinical study performed at our laboratory comparing color Doppler grading of AI in 29 patients with the current gold standard, angiography.21 This same feature (proximal width) was found to be the most reliable predictor of the angiographic grade of AI severity.

Both the laser and continuous-wave Doppler studies showed that the jets from the largest defect (5 mm) had their greatest velocity in early diastole, whereas the velocity of jets from smaller defects (3 and 1 mm) did not significantly decrease throughout diastole. Similarly, both color Doppler and streak photography methods showed that large defects produced jets with variant (disturbed) flow patterns early in diastole with laminar (monochromatic) flow occurring later in diastole, whereas small defects had significant color variance throughout their duration (figure 3, C and D). These features should also be considered when grading AI jets.

Therefore, analysis of color Doppler jet morphology for grading severity of AI needs to be seriously reconsidered. The primary determinate used for grading AI should be the proximal width of the jet in the immediate subvalvar region, proximal to the point where the jet begins to expand to its maximal downstream width (which is also pressure dependent). Depending on the instantaneous loading conditions, the maximal jet depth alone will often give misleading grading information. The most important hemodynamic factor influencing AI jet morphology is the pressure gradient. This ΔP may be at least estimated in patients with AI by measurement of the mean and diastolic blood pressures. Left ventricular diastolic pressure can not be determined noninvasively, but it may be at least estimated clinically, as well as derived from the continuous-wave Doppler determination of the diastolic gradient across the aortic valve. With knowledge of the estimated ΔP, supplemental grading information may be obtained from the jet's maximal length, maximal downstream width, its area, and the relative duration of its disturbed (variant) flow. We therefore strongly recommend that all Doppler evaluations of AI include a simultaneous blood pressure determination. Only then may the jet depth, maximal width, and area be appropriately interpreted. For similar reasons, this applies equally to mitral incompetence.

This information is equally relevant for the techniques used for mapping and grading of AI by a pulsed Doppler method. This has traditionally been quantitated as a function of the maximal depth to which an AI jet can be detected. Misleading conclusions will therefore commonly arise when extrapolating such measurements of jet depth (or velocity) to predictions of the valve's regurgitant volume or defect size.

**Mechanical valves.** For normal mechanical valves, we found that the physiologic incompetence related to closure of the leaflets was limited to very early diastole. Its duration was dependent on loading conditions and showed an inverse relationship to either the MAP (maximal when <125 mm Hg) or cardiac output (table 3). This relationship has been previously demonstrated in a pulse duplicator system.22 However, the jets from mechanical valves with pathologic incompetence (induced by a mock thrombus) were pendiastolic and became both deeper and wider with an increased MAP. Hence, the color jets related to physiologic and pathologic AI from mechanical valves behave very differently, both in the baseline state and during extreme loading conditions.

**Limitations.** We found that color Doppler imaging has excellent potential resolution, at least in this idealized system in vitro. Flow was detectable with a cardiac output as low as 0.5 liter/min, AI could be demonstrated even through a 1 mm pinhole (depending on the hemodynamic variables), and the lateral resolution allowed distinction of two jets originating only 3 mm apart. However, equally sensitive resolution would not be expected in most patients because of the acoustic limitations imposed by closed-chest imaging, and because of the greater distances involved.

We used circular valvular defects. The irregular margins of a regurgitant orifice in vivo may produce a somewhat variable proximal width in different planes. Such measurements obtained in vivo should therefore include multiple imaging planes of the proximal jet in the left ventricular outflow tract.

Furthermore, the left ventricular outflow tract in our system in vitro was circular and its diameter was fixed at 25 mm. The potential influence of a dynamic outflow tract (having a complex shape and enlarging throughout diastole) on AI jet morphology remains to
be further defined with measurements in vivo in open-chest dogs. Our experience comparing the proximal width of AI jets to angiographic grading in patients has shown that this is a valuable Doppler grading system clinically, and is most specific when the proximal width of the jet is corrected for the diameter of the left ventricular outflow tract.21

In conclusion, we have found that the morphology of AI jets as imaged by color Doppler flow mapping is profoundly dependent not only on the RF, but also on the defect size and the instantaneous hemodynamic conditions. Use of the jet's maximum length or width as independent predictors of the severity of AI is not a reliable method. These dimensions are primarily determined by velocity and are often not proportional to the regurgitant volume. The minimal proximal jet width immediately below the valve plane is the most reliable independent predictor of both defect size and the severity of AI (the RF). Other variables (including jet length, width, and duration of color variance) provide supplemental grading information, but only at a known (noninvasively approximated) ΔP.

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