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SEMINAR ON IN VITRO STUDIES OF CARDIAC FLOW AND THEIR APPLICATIONS FOR CLINICAL DOPPLER ECHOCARDIOGRAPHY—VI*

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Factors Influencing the Structure and Shape of Stenotic and Regurgitant Jets: An In Vitro Investigation Using Doppler Color Flow Mapping and Optical Flow Visualization

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To evaluate factors influencing the structure and shape of stenotic and regurgitant jets, Doppler color flow mapping and optical flow visualization studies were performed with use of a syringe model with a constant rate of ejection to simulate jets of valvular regurgitation and a pulsatile flow model of the right heart chambers to simulate jets of mild, moderate and severe valvular pulmonary stenosis. Ink-(0 to 40%) glycerol-water jets (viscosity 1 to 3.5 centiPoise) were produced by injecting the fluid at a constant rate into a 10 gallon rectangular reservoir of the same still fluid through 1.4 and 3.4 mm needles. The Doppler color flow scanners imaged the laminar jet length within 3 mm of actual jet length (2 to 6 cm) and the jet width within 2 to 3 mm of the actual jet width. Jet flows with Reynolds numbers ranging from 230 to 1,200 injected into still fluid yielded jet length/width ratios that decreased with increasing Reynolds numbers and leveled off to a length/width ratio of 5–6:1 at a Reynolds number near 600. When the fluid reservoir was swirled to better mimic the effect of flow entering the same cardiac chamber from a second source, the jets showed diminution of the jet length/width ratio and a clearly defined zone of turbulence.

Studies of the pulsatile flow model were performed at cardiac outputs of 1 to 6 liters/min for the normal and each stenotic valve. Mild stenosis had an orifice area of 2.8 cm^2, moderate stenosis an area of 1.0 cm^2 and severe stenosis an area of 0.5 cm^2. Laminar jet length represented the length of the total jet, which had a symmetric width and was measured from the valve opening to a region where the jet exhibited a spray effect. Laminar jet lengths (0.2 to 1.1 cm) were imaged by Doppler color flow mapping and optical visualization only in the moderate and severely stenotic valves and only at flows $\leq$3 liters/min (mean Reynolds numbers $\leq$3,470). Beyond this flow rate the jets exhibited a spray effect. Laminar jet length/width ratio approached unity with an increased amount of valvular stenosis and higher flow volumes (cardiac output). Proximal aliasing was present in each valve studied. The length of aliasing (0 to 3.2 cm) proximal to the valve was longer with increased flow rates and increased amounts of stenosis.

In summary, the structure and shape of jets were dependent on the orifice diameter, driving pressure (cardiac output), fluid viscosity and presence or absence of radial flow velocities that interact with the jet.

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High velocity flow through obstructive orifices creates jets that are a frequently observed phenomenon in cardiology. Jets may be produced by stenotic or regurgitant valves, or both, stenotic arteries, stenotic synthetic grafts (for example, Blalock-Taussig shunts, systemic to pulmonary artery

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shunts and coronary artery bypass grafts) and restrictive atrial or ventricular communications. These jets may be visualized noninvasively by Doppler color echocardiography (1–10) and invasively by angiography. Stenotic laminar jets in an unbounded state have been found to follow a relation between laminar jet length and width: as the stenosis becomes more severe and approaches a Reynolds number >800, the laminar jet length decreases but the ratio of the laminar jet length to width remains near 5:1 (11). Wranne et al. (12) found that the absolute distance of the intrusion of the jet or jet length was dependent on the volume flow through the orifice and the driving pressure. Jet length is longer for increased flow volumes and higher driving pressures. Doppler color flow mapping studies performed on patients with valvular disease have rarely shown such long narrow jets.

Bird et al. (13) have shown that there is a subvalvular pressure gradient in a region called the vena contracta that is demonstrable in isolated aortic stenosis. The same phenomenon may be observed with regard to velocity acceleration with pulsed and Doppler color flow mapping. This study was designed to investigate the formation and shape of jets produced in vitro by a syringe with various needle diameters and a pulsatile flow model of the right heart chambers. The syringe needle model was used to simulate jets created by valvular regurgitation. The pulsatile right heart model was used to simulate jets produced by pulmonary stenosis. Jets were imaged by optical flow visualization and Doppler color flow mapping. The observations obtained may help explain clinically observed findings in the practice of noninvasive cardiology.

Methods

Laminar jets produced by syringe. Laminar jets were produced by injecting a solution of ink, 0 to 40% glycerol and water at a constant rate of 1 ml/s, through a syringe needle with an internal diameter of either 1.4 or 3.4 mm, into a 10 gallon rectangular (52 cm long × 28 cm wide × 36 cm tall) reservoir of the same still fluid. The fluid viscosity ranged from 1.0 to 3.5 centiPoise (cP). By varying the fluid viscosity and needle diameter, the Reynolds number (ratio of inertial and viscous forces) was varied over a range of 230 to 1,200. Reynolds number was chosen as the varied parameter because it characterized a flow field as laminar, transitional or turbulent. The rate of injection was measured by an electromagnetic flow meter and by continuous wave Doppler ultrasound. The jets were imaged by optical visualization with a video camera and by Doppler color flow mapping ultrasonography with use of a 5 MHz transducer. The transducer was held steady and the syringe pointed toward or away from the footprint so that the flow was optimized parallel to the beam of the sound waves. Three Doppler color flow scanners were utilized: Aloka 880, Toshiba SSH65A and Hewlett-Packard prototype 77020. The instruments were run at a low flow gain and the lowest available pulsed repetition frequency. Jets were recorded on videotape for later analysis at slow speed.

To better mimic the jets seen in patients with valvular cardiac disease, the fluid in the reservoir was swirled with a magnetic stirrer. To mimic the effect of mitral inflow on aortic regurgitation, the fluid was swirled in the direction of the jet. To mimic the effect of pulmonary venous inflow on mitral regurgitation, the fluid bath was swirled against the direction of the jet.

The videotapes were reviewed to allow measurement of laminar jet length and width from both the optically recorded jets and the Doppler color flow jets. Laminar jet length was measured from the needle tip to an area where the jet broke up into turbulence and developed a spray effect. The laminar jet length was characterized by a uniform jet width. Laminar jet width was measured at the needle orifice (Fig. 1). A minimum of 12 jets were measured under each condition and a mean value ± SD was obtained. Reynolds number (Re) for the various conditions was calculated with use of the formula:

$$\mathrm{Re} = \frac{\text{mean flow velocity} \times \text{orifice diameter}}{\text{fluid kinematic viscosity}}.$$

Pulsatile flow apparatus. The pulsatile flow apparatus was set up to duplicate the flow and pressure wave forms on the right side of the human heart. It consisted of a physiologically shaped plexiglass model of the pulmonary artery into which varying pericardial bioprosthetic valves were inserted at the pulmonary valve area. Stenosis was created by sewing the valve leaflets together with polyester thread. A centrifugal pump and plastic bucket were used to fill and empty the pulse duplicator. A series of pressure taps, valves and pressure reservoirs were designed to simulate the right heart. A detailed description and illustration of the design and function of the pulse duplicator and the pulmonary artery model have been published elsewhere (14).

The right heart pulse duplicator was operated under the following physiologic conditions: 1) heart rate of 70 beats/min; 2) systolic time interval of 320 to 350 ms; 3) mean pulmonary artery pressure of 15 to 30 mm Hg; and 4) cardiac output varying from 1 to 6 liters/min. An aqueous glycerine solution containing 2% cornstarch particles (10 μm diameter) was used as the blood analog fluid and adjusted to a physiologic viscosity of 3.5 cP. The normal pulmonary valve was a 29 mm pericardial valve with a valve area of 5.3 cm^2 and radius of 1.3 cm. Mild, moderate and severe pulmonary stenosis was created by sewing together the valve leaflets of three 25 mm pericardial trileaflet valves. The valve areas were determined by assuming the orifice to be a circle and calculating the area to be πr^2, where r = the radius of the open valve as measured with a ruler. The respective pulmonary valve areas were 2.8, 1.0 and 0.5 cm^2 (radius = 0.9, 0.6 and 0.4 cm, respectively) with resultant respective peak pressure gradients of 13.9, 58.9 and 97.5 mm Hg at a cardiac output of 6 liters/min. The pulmonary artery bifurcation was made to geometric dimensions that had been previously obtained from detailed measurements of pulmonary angiograms and autopsy specimens. The combination of the physiologically realistic pulsatile flow system and bioprosthetic valves make this model resemble the physiology encountered in forms of congenital and acquired valvular heart disease.

Jets produced by the various pulmonary valves were imaged through an optical flow visualization setup and by Doppler color flow ultrasonography and recorded on videotape for analysis. For each valve area and cardiac output, the laminar jet length and width were measured 15 times by a single observer (K.A.K.) who was aware of the variables. A mean value ± SD was calculated for each condition (Fig. 2). Laminar jet length was measured from the valve opening to the area where the jet became more turbulent and developed a spray effect; jet width was measured just beyond the valve orifice for optical flow visualization studies. As defined, the laminar jet length constitutes a small portion of the total jet length. Upstream mean Reynolds numbers (Re) were calculated with use of the peak flow across the pulmonary valve with the equation:

$$\text{Mean Re} = \frac{\text{fluid density} \times \text{mean velocity} \times \text{vessel diameter}}{\text{fluid viscosity}}.$$

The fluid density was 1.05 g/cm^3; vessel diameter was 3.10 cm and fluid viscosity was 3.5 cP.

Doppler color flow mapping studies of pulsatile model. The pulmonary artery model had been built with a port proximal to the insertion of the bioprosthetic valve that allowed one to perform two-dimensional and Doppler color flow mapping studies of the model with the direction of flow parallel to the ultrasound beam. Jets produced by the pulsatile pump were imaged by an Aloka 880 scanner with a 2.5 MHz phased array transducer with a Nyquist limit of 0.92 m/s. The color gain was set and maintained at 5. The Doppler color flow map was used to delineate the pattern of pulsatile flow proximal and distal to the various pulmonary valves. Specific variables of interest were: 1) length of aliased color flow proximal to the valves; 2) width of the jet produced distal to the valve measured at the level of the leaflet opening; and 3) length of the laminar jet produced by the stenotic valve (the jet was measured from the valve opening plane to the area where it became more turbulent or exhibited a spray effect). Measurements were made at peak systole for 15 jets and a mean value ± SD calculated. One observer (K.A.K.) made the measurements with the knowledge of the valve area and volumetric flow. This same observer measured the optical flow values; hence, there may be potential bias when instrument and Doppler to optical visualization comparisons were made.

Results

Laminar syringe needle jets. Laminar jet lengths produced by a 1.4 mm needle varied between 1.9 ± 0.2 to 6.1 ± 0.6 cm in the 0 to 40% glycerine-water solution with corresponding Reynolds numbers of 350 to 1200 (Table 1). Laminar jet lengths did not vary between measurements from jets recorded by optical visualization and any of the three Doppler color flow scanners. A close correlation with an r value of 0.94 was found by plotting the mean Doppler color flow jet length against the mean optical visualization jet length (Fig. 3). Jet width was similar among the three Doppler color flow scanners; however, the scanners overestimated the true jet width by approximately 3 mm because of problems with lateral resolution and jet-entrained flow. Laminar jet length to width ratio decreased from 37.3 ± 1.1 to 13.9 ± 0.6 with increasing Reynolds numbers; however, a constant value was not approached (Fig. 4). The total jet length was not measured, only the portion that was defined as laminar in the Methods section.

By utilizing a larger needle with an internal diameter of 3.4 mm, the laminar jet lengths varied from 1.0 ± 0.1 to 7.3 ± 0.7 cm over a range of Reynolds numbers from 200 to 1,200 (Table 2). Only the optical visualization study was undertaken because the Doppler color flow studies correlated well in the 1.4 mm needle experiment. Laminar jet length to width ratio approached a value of 5–6:1 at a Reynolds number near 600 (Fig. 5). At larger Reynolds numbers, the length to width ratio decreased minimally.

Doppler color flow studies in patients with a regurgitant or stenotic valve or both, have rarely shown such long, narrow jets as those produced by these syringe needle experiments. In addition, there was no variance encoding (green color) by the flow scanners in the distal portion of the jet, which is known to have low velocity turbulence. To mimic the effect of radial velocities on laminar jets, the fluid bath was swirled either toward or against the direction of the laminar jet. When the bath was swirled against the direction of the laminar jet, the jet length/width ratio decreased markedly toward unity (Fig. 6) and the jet shape was no longer linear but curvilinear, making measurements of jet length difficult. In addition, a turbulent region was clearly identifiable with the Doppler color flow map in the region where it was not encoded in the still fluid experiment. When the fluid bath was swirled in the direction of the laminar jet, the same phenomena were observed.

An area of blackout was present in the middle of some laminar jets imaged by the Toshiba Doppler color flow scanner. This area represented a loss in color encoding (Fig. 1C) and was not seen in the images recorded by the Aloka or the Hewlett-Packard scanners.

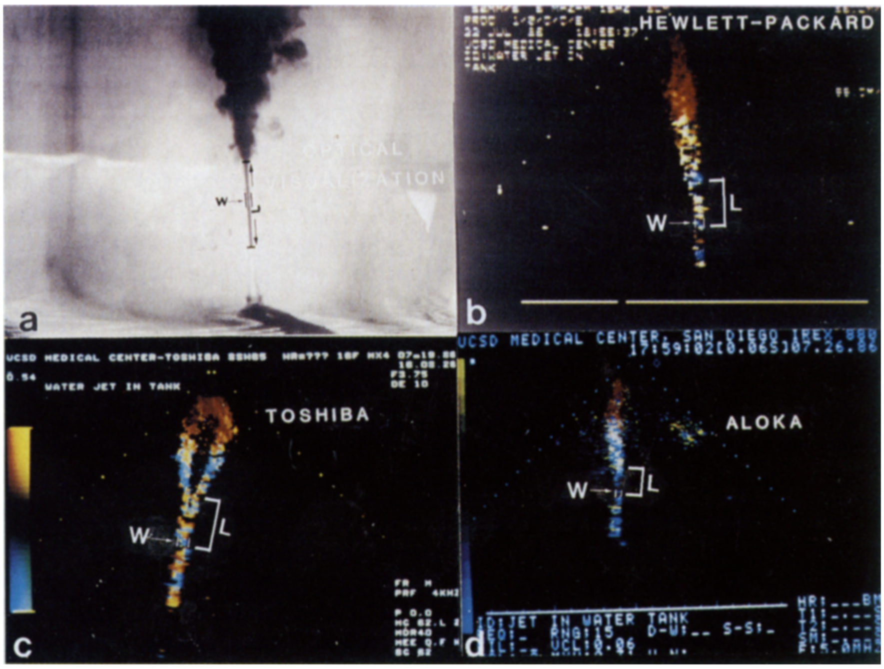


Figure 1. Illustrations of laminar jets as imaged by optical flow visualization (**a**) and by the Hewlett-Packard prototype 77020 (**b**), Toshiba SSH65A (**c**) and Aloka 880 (**d**) Doppler color flow scanners. Measurement of the laminar jet length (L) was from the end of the needle to a point where the jet dispersed into low velocity turbulence as marked in the illustration by the lines. The jet width (W) was measured at the needle tip. Note that the laminar jet length represents a small portion of the total jet length.

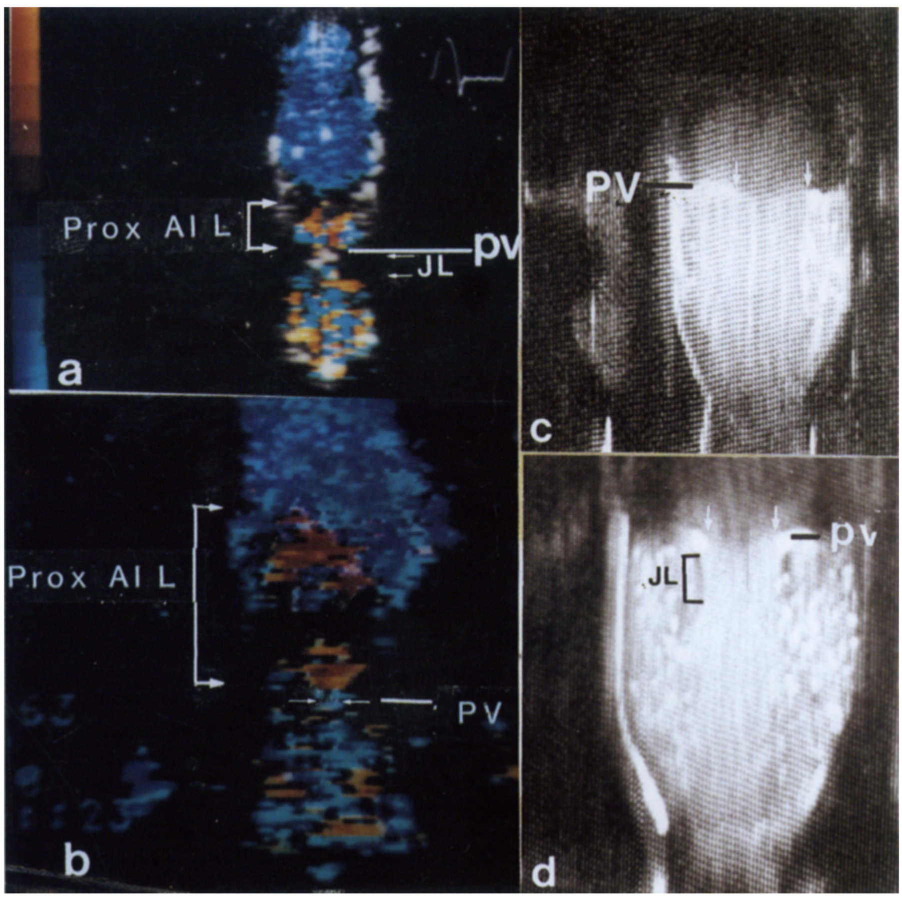


Figure 2. Illustrations of the jets produced by the pulsatile flow model. The direction of flow was away from the transducer. **a** and **b** illustrate Doppler color flow imaged stenotic jets under two different conditions. Panel **a** illustrates a jet that has a short laminar length (JL) and a zone of proximal aliasing (Prox Al L). In contrast, panel **b** illustrates a stenotic jet that had a spray effect and no true laminar length. The level of the pulmonary valve (PV) is denoted with a **bold line**. The jet width was measured between the **small horizontal white arrows**. The length of proximal aliasing (Prox Al L) shown was longer than that in panel **a**. Panels **c** and **d** are examples of optically visualized flow profiles through the pulmonary valve (PV) under different experimental conditions from panels **a** and **b**. Panel **c** represents the optical flow visualization of a mildly stenotic pulmonary valve that exhibited a streamlined flow profile with no laminar jet flow, whereas panel **d** illustrates the type of laminar jet lengths observed. The laminar jet length (JL) has been measured between the **black lines**, whereas the jet width was measured between the **small vertical arrows**.

Table 1. Laminar Jet Lengths Produced By a 1.4 mm Needle and Measured Over a Range of Reynolds Numbers

% Glycerine-Water	Reynolds Numbers	Laminar Jet Length (cm) Opt. Vis.	Aloka	Toshiba	Hewlett-Packard	Laminar Jet L/W*
0	1,200	1.9 ± 0.2	2.3 ± 0.4	2.3 ± 0.6	1.8 ± 0.3	13.9 ± 0.6
15	780	2.2 ± 0.2	2.6 ± 0.4	3.0 ± 0.3	ND	16.1 ± 0.9
25	580	4.2 ± 0.4	3.6 ± 0.4	3.8 ± 0.5	ND	28.2 ± 0.6
40	350	5.8 ± 0.7	6.1 ± 0.6	5.9 ± 0.5	6.3 ± 0.8	37.3 ± 1.1

*Calculated from jet lengths and widths obtained with optical visualization data. L/W = length/width ratio; ND = no data available; Opt. Vis. = optical visualization; W = width.

Jets Produced by the Pulsatile Flow Model

Doppler color flow maps of stenotic jets. For the normal pulmonary valve study, the flow that emerged from the valve was evenly distributed in the main pulmonary artery and no jet type of flow field was observed. Laminar Doppler color flow jets were produced by the valves exhibiting mild (valve area = 2.8 cm^2), moderate (valve area = 1.0 cm^2) and severe pulmonary stenosis (valve area = 0.5 cm^2) (Table 3).

For mild pulmonary stenosis, laminar Doppler color flow jets were produced only at a volumetric flow rate of 3 liters/min (mean Reynolds number 3,470) with laminar jet length of 5.3 ± 1.3 cm; whereas the jet width was 1.8 ± 0.1 cm yielding a jet length/width ratio of 2.9. At higher cardiac outputs or larger mean Reynolds numbers, a true laminar jet was no longer observed but rather a funnel-shaped jet spray effect was seen. For smaller volumetric flow rates the flow across the valve was streamlined.

For moderate pulmonary stenosis, laminar Doppler color flow jets were produced at a volumetric flow rate of 1.0 liter/min (mean Reynolds number = 1,160) and remained laminar up to a flow rate of 3.0 liters/min. The laminar jet length varied between 1.1 ± 0.1 to 0.5 ± 0.2 cm, whereas the jet width ranged from 1.0 ± 0.1 to 1.1 ± 0.2 cm, thus yielding laminar jet length/width ratios of ≤1.0.

In severe pulmonary stenosis, laminar Doppler color flow jet length varied between 1.1 ± 0.1 to 0.2 ± 0.1 cm, whereas the jet width ranged from 0.8 ± 0.1 to 0.7 ± 0.1 cm at volumetric flow rates of 1 to 2 liters/min (mean Reynolds numbers of 1,160 to 2,310). Flow rates >2 liters/min produced a funnel-shaped jet spray effect, rather than a laminar jet that broke up into a spray effect. The jet length/width ratio ranged from 1.4 downward.

In all three models of pulmonary stenosis, the Doppler color flow measured jet width was within 0.2 cm of the known diameter of the valve (mild stenosis = 1.8 cm, moderate stenosis = 1.2 cm and severe stenosis = 0.8 cm).

Optical flow visualization of stenotic jets. A similar phenomenon was observed in the optically visualized jets (Table 4). Laminar jets were not produced by the normal and mildly stenotic pulmonary valves. Moderate pulmonary stenosis created laminar jets whose jet length varied from 1.2 ± 0.1 cm to 0.5 ± 0.1 cm for volumetric flow rates of 2 to 3 liters/min. A funnel-shaped jet spray that had no true laminar jet length was seen at flow rates >3 liters/min. The jet width at the valve opening ranged from 0.8 ± 0.1 to 1.0 ± 0.1 cm. The

Figure 3. Graph of optically visualized jet length (JL) versus Doppler color flow imaged jet length (L) measured in centimeters. The line of identity **(dotted black line)** was plotted to allow for comparison. A close correlation was found (correlation coefficient = 0.94).

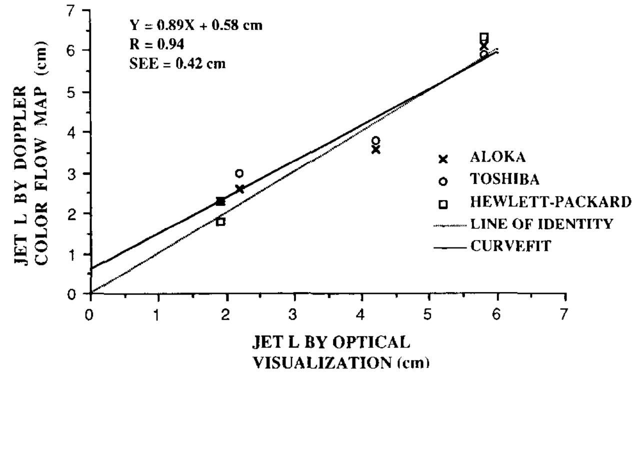


Figure 4. Graph illustrating the relation between Reynolds number (a measure of peak flow and pressure head) and laminar jet length/width ratio (L/W) for the needle orifice of 1.4 mm.

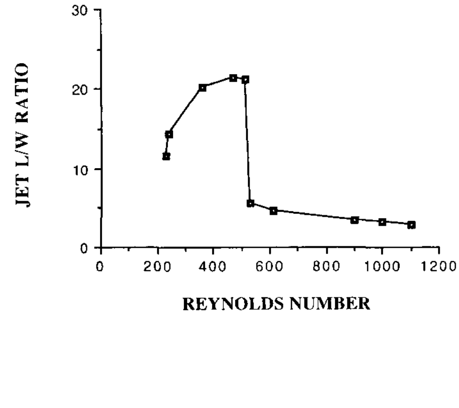


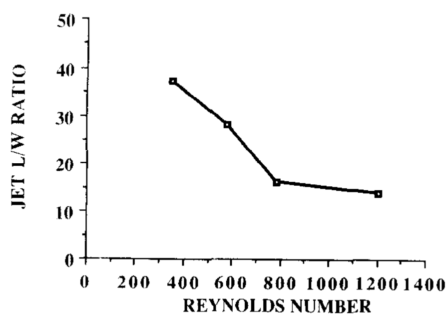
Table 2. Laminar Jet Length (L) and Width (W) Produced By a 3.4 mm Needle and Measured Over a Range of Reynolds Numbers

% Glycerine-Water	Reynolds Number	Jet L (cm)	Jet W (mm)	Jet L/W
0	1,100	1.0 ± 0.1	3.4	3.0
0	1,000	1.1 ± 0.1	3.4	3.3
0	900	1.2 ± 0.2	3.4	3.6
15	610	1.6 ± 0.1	3.4	4.7
15	530	1.9 ± 0.2	3.4	5.6
25	510	7.2 ± 0.3	3.4	21.2
25	470	7.3 ± 0.5	3.4	21.4
40	360	6.9 ± 0.6	3.4	20.2
40	240	4.9 ± 0.5	3.4	14.4
40	230	4.0 ± 0.5	3.4	11.7

resultant jet length/width ratio varied from 1.5 downward. Severe pulmonary stenosis produced a funnel-shaped jet spray effect at all the volumetric flow rates studied.

Proximal aliasing seen by Doppler color flow maps in stenotic jets. Proximal aliasing was evaluated by Doppler color flow mapping (Table 5, Fig. 2). The normal pulmonary valve first exhibited proximal aliasing at a volumetric flow rate of 5.0 liters/min. The average length of aliasing was 1.4 ± 0.2 cm proximal to the plane of the pulmonary valve opening during peak systole. At the highest flow rate evaluated, 6.0 liters/min, aliasing was seen 1.8 ± 0.2 cm proximal to the valve. Mild pulmonary stenosis created aliasing that was first observed 0.5 ± 0.1 cm proximal to the valve at a flow rate of 3.0 liters/min; the length of proximal aliasing increased to a distance of 2.1 ± 0.2 cm at a flow rate of 6.0 liters/min. Proximal aliasing of 0.3 ± 0.1 cm was seen for the moderately stenotic pulmonary valve at a volumetric flow rate of 2.5 liters/min and increased to 3.2 ± 0.2 cm at a flow rate of 6.0 liters/min. The severely stenotic pulmonary valve created aliasing that was initially seen 0.7 ± 0.4 cm proximal to the valve at a flow rate of 2.0 liters/min. When the flow rate was 6.0 liters/min, the aliasing extended 3.2 ± 0.2 cm proximal to the valve. With increasing amounts of stenosis and higher volumetric flow rates or mean Reynolds numbers, a longer length of proximal aliasing was observed (Fig. 7).

Figure 5. Graph illustrating the relation between Reynolds number and laminar jet length/width (L/W) ratio for the needle orifice of 3.4 mm.



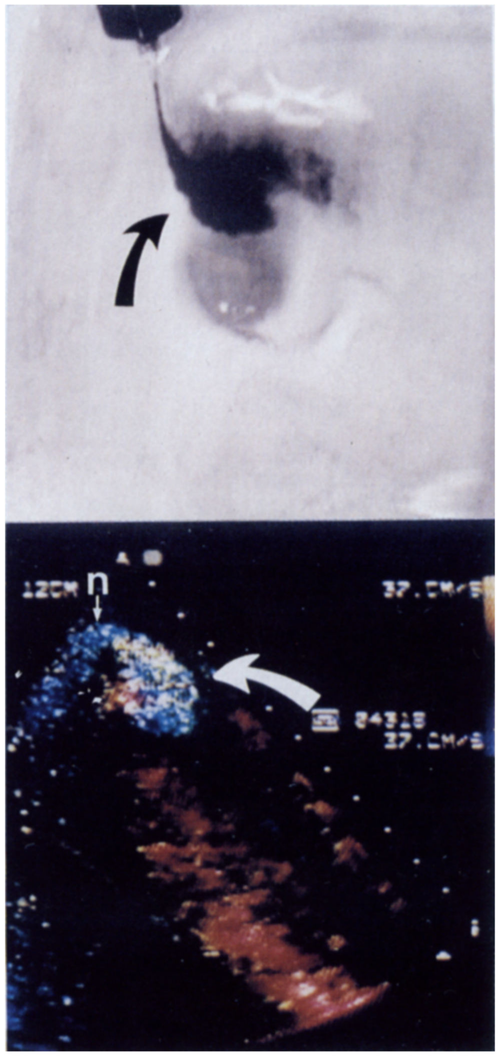


Figure 6. Illustration of the effect of swirling on laminar jets produced through a needle orifice. The **top panel** represents the optically visualized jet. The **large black arrow** shows the direction in which the reservoir fluid was swirling. Note the shorter length of the laminar jet from Figure 1. The **bottom panel** represents the jet imaged by the Hewlett-Packard Doppler color flow scanner. The **large white arrow** represents the direction of the swirling reservoir fluid and n, the needle tip. Note the shorter laminar jet length and the presence of variance within the dispersed portion of the jet.

Discussion

Multiple factors have been found that influence the structure and shape of unbounded (regurgitant) and stenotic jets in two in vitro flow systems. Jets produced by injecting a 0 to 40% glycerine-water fluid into a large reservoir of the

Table 3. Doppler Color Flow Laminar Jet Length (L) and Width (W) Measured At Various Mean Reynolds Numbers in the Pulsatile Flow Model of the Right Heart

Reynolds Numbers	Volumetric Flow (liters/min)	Mild PS			Moderate PS			Severe PS		
		Jet L	Jet W (cm)	Jet L/W	Jet L	Jet W* (cm)	Jet L/W	Jet L	Jet W* (cm)	Jet L/W
1,160	1.0		No jet observed		1.1 ± 0.1	1.1	1.0	1.1 ± 0.1	0.8	1.4
1,610	1.5				1.0 ± 0.1	1.1	1.0	0.8 ± 0.1	0.7	1.1
2,310	2.0		No jet observed		0.7 ± 0.2	1.1	0.6	0.2 ± 0.1	0.7	0.3
2,890	2.5				0.8 ± 0.1	1.1	0.7	Jet spray	0.7	†
3,470	3.0	5.3 ± 0.1	1.8 ± 0.1	2.9	0.5 ± 0.1	1.0	0.5	Jet spray	0.7	†
3,690	3.5				Jet spray	1.1	†	Jet spray	0.7	†
3,910	4.0	Jet spray	1.9 ± 0.2	†	Jet spray	1.2	†	Jet spray	0.7	†
4,150	4.5				Jet spray	1.1	†	Jet spray	0.7	†
4,370	5.0	Jet spray	1.8 ± 0.1	†	Jet spray	1.2	†	Jet spray	0.7	†
4,550	5.5				Jet spray	1.1	†	Jet spray	0.8	†
4,610	6.0	Jet spray	1.8 ± 0.1	†	Jet spray	1.2	†	Jet spray	0.8	†

*Standard deviation for all means = 0.1 except for the first one under moderate PS of 0.2; †Jet L/W approaches 0. PS = pulmonary stenosis; other abbreviations as in Table 1.

same still and swirling fluid allowed the evaluation of factors governing the structure and shape of unbounded jets. In contrast, the pulsatile flow model allowed the investigation of factors governing the formation of jets through stenotic valves. The pulsatile flow apparatus used in the latter investigations had various control features that permitted accurate control of the fluid mechanics duplicating the right side of the human heart, thereby making it possible to obtain highly repeatable settings so that the four pulmonary valves could be studied at identical flow conditions. As in other studies utilizing the pulsatile flow model (14–17), the use of bioprosthetic pericardial trileaflet valves was chosen because, when sutured at the commissures to restrict the effective flow area, they act like stenotic pulmonary valves in humans. A 2% cornstarch-glycerine-water solution acted as the blood analog because preliminary studies indicate no difference between this solution and erythrocytes fixed in 50% glutaraldehyde (14). Glutaraldehyde-fixed erythrocytes are the closest long-lasting and reusable analog to in vivo erythrocytes for ultrasound studies (18).

Laminar syringe needle jets. Laminar jets produced in an unbounded state represented an idealized model of valvular regurgitation. Caution, however, must be exercised in extrapolating these in vitro data to in vivo clinical situations of valvular regurgitation because the model is different from reality. First, the model utilized constant rather than pulsatile flow. Second, the cardiac chamber receiving the regurgitant jet may bound it, whereas the fluid reservoir did not. Third, the rate and volume of flow through the needle were lower than those often encountered clinically.

Wranne et al. (12) found that the length of the laminar jet was linearly related to the mean velocity of the fluid in the orifice multiplied by the orifice diameter. However, if one tried to correlate the jet length to regurgitant volume, one also had to account for the flow area of regurgitant flow. The results of our initial study that produced unbounded jets through a syringe demonstrated that the ratio of laminar jet length to width decreased with increasing Reynolds number by varying either the fluid viscosity or the orifice diameter through which the fluid was ejected. Changing the diameter of the needle changed the volumetric flow area. The ratio of the laminar jet length to width varied from 5:1 to 40:1 over a range of Reynolds numbers from 200 to 1,200. At a Reynolds number of approximately 600, the laminar jet length to width

Table 4. Optically Visualized Laminar Jet Length (L) and Width (W) Measured at Various Mean Reynolds Numbers in the Pulsatile Flow Model of the Right Heart

Reynolds Numbers	Volumetric Flow (liters/min)	Mild PS			Moderate PS			Severe PS		
		Jet L	Jet W (cm)	Jet L/W	Jet L	Jet W* (cm)	Jet L/W	Jet L	Jet W* (cm)	Jet L/W
2,310	2.0		No jet observed		1.2 ± 0.1	0.8	1.5	Jet spray	0.9	†
3,470	3.0		No jet observed		0.5 ± 0.1	0.9	0.6	Jet spray	0.9	†
3,910	4.0		No jet observed		Jet spray	0.9	†	Jet spray	0.9	†
4,370	5.0		No jet observed		Jet spray	0.9	†	Jet spray	0.9	†
4,610	6.0		No jet observed		Jet spray	1.0	†	Jet spray	0.9	†

*Standard deviation for all the means = 0.1; †approaches 0. PS = pulmonary stenosis; other abbreviations as in Table 1.

Table 5. Distance of Aliasing Proximal to Pulmonary Valve Measured by Doppler Color Flow at Various Mean Reynolds Numbers in the Pulsatile Flow Model of the Right Heart

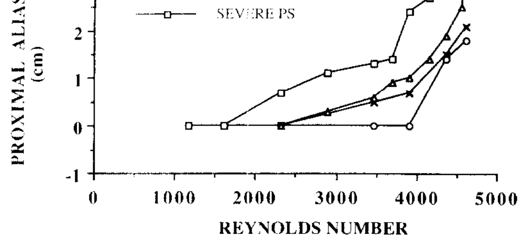
Reynolds Numbers	Volumetric Flow (liters/min)	Proximal Aliasing (cm) Normal PV	Mild PS	Moderate PS	Severe PS
1,160	1.0	ND	ND	0.0	0.0
1,610	1.5	ND	ND	0.0	0.0
2,310	2.0	0.0	0.0	0.0	0.7 ± 0.4
2,890	2.5	ND	ND	0.3 ± 0.1	1.1 ± 0.1
3,470	3.0	0.0	0.5 ± 0.1	0.6 ± 0.2	1.3 ± 0.1
3,690	3.5	ND	ND	0.9 ± 0.2	1.4 ± 0.2
3,910	4.0	0.0	0.7 ± 0.1	1.0 ± 0.1	2.4 ± 0.3
4,150	4.5	ND	ND	1.4 ± 0.3	2.7 ± 0.2
4,370	5.0	1.4 ± 0.2	1.5 ± 0.1	1.9 ± 0.3	2.9 ± 0.2
4,550	5.5	ND	ND	2.5 ± 0.3	3.0 ± 0.1
4,610	6.5	1.8 ± 0.2	2.1 ± 0.2	3.2 ± 0.2	3.2 ± 0.2

ND = no data available; PS = pulmonary stenosis; PV = pulmonary valve.

ratio leveled off at 5 to 6:1 for a needle diameter of 3.4 mm. This same phenomenon was not evident for the 1.4 mm needle, presumably due to the fact that jets were not produced at Reynolds numbers <350 or >1,200. In the clinical situation, changes in volumetric flow rates and pressure heads would be more likely to result in changes in Reynolds numbers and mean velocities rather than changes in fluid viscosity or orifice diameter. Although we did not evaluate the effect of the volumetric flow through the needle, it has been shown that volumetric flow and mean flow velocity are directly related to the Doppler color flow area produced by the dissipating turbulent jet spray and its length, whereas laminar jet width had a minimal influence on the Doppler color flow area (19). Hence, the length of laminar jets in still fluid was dependent on the orifice diameter, fluid viscosity, volumetric flow, pressure head and other factors influencing Reynolds number.

Regurgitant jet volumes measured for unbounded jets by Wranne et al. (12) were larger than those observed clinically.

Figure 7. Graph illustrating the effect of Reynolds number on the amount of proximal aliasing in centimeters for the four pulmonary valves (PV) studied. PS = pulmonary stenosis. More proximal aliasing was seen with higher Reynolds numbers representing larger cardiac outputs and with more stenotic valves.



Roschke and Back (11) found that the jets created in vitro in an unbounded state were larger than those seen in vivo, presumably because of the more complex geometry found in vivo. Davidoff et al. (20) attempted to integrate jet volumes from Doppler color flow maps, and their results exceeded the actual regurgitant volumes. They believed that regurgitant volumes cannot be accurately measured by Doppler color flow maps. They assumed the jets were symmetric; however, in reality the jets not only may be swirling but may be swirling assymmetrically. In our model, when the reservoir fluid was swirled, there was diminution of the laminar jet length/width ratio, deceleration and spreading of the jet on Doppler color flow mapping and a clear transition to turbulence into the expanded jet. Hence, the structure and shape of jets were also dependent on the radial flow velocities that interact with them, such as other flows entering the same chamber. This is an important clinical concept when using Doppler color flow mapping to estimate the amount of valvular regurgitation.

Low velocity turbulence was present in the expanded portion of the laminar jet. Review of the videotapes of the Doppler color flow imaged jets failed to show any green color, an indicator of variance, in that region for jets produced in still fluid. It was only when the fluid reservoir was swirled that variance was encoded by the Doppler color flow scanners. Hence, higher velocity turbulence was created by laminar jets interacting with radial velocities. Low velocity turbulent flow had less percentage variability compared with the available velocity scale and encoded variance less than higher velocity turbulent flow.

The artifact noted on some Toshiba flow scanner studies was an area of blackout in the middle of the laminar and expanded jet (Fig. 1c). When the jets were produced, microcavitations were created by the injections that were imaged as a tissue echo of low echogenicity because the Toshiba scanner had a tissue pixel priority algorithm. The scan converter algorithm does not encode flow velocities as color onto pixels already encoded with a tissue signal so that no color flow was encoded in the middle of the jet. This artifact was not present in the Aloka- or Hewlett-Packard–imaged Doppler color flow jets. We have not as yet had a problem with this phenomenon clinically, but it may exist in patients whose blood is echogenic such as that of patients whose blood flows through certain abnormal mechanical valve prostheses.

Exact results from our syringe study or similar ones must not be directly extrapolated to reality. In addition to the reasons cited previously, other conditions make these jets unlike those seen from regurgitant valves: 1) the system employed has steady rather than pulsatile flow; 2) the influence and elasticity of the heart walls and blood vessels were ignored; 3) the system assumed that the orifice was symmetric and that its diameter remained constant during the regurgitant or stenotic flow period; 4) it neglected the effects of bends, branches and bifurcations on jet formation; and 5) the rate and volume of flow through the needle orifice may not represent that found clinically. The evaluation of stenotic jets produced by the right heart pulse duplicator overcame some of these limitations.

Stenotic jets produced by pulsatile flow apparatus. Stenotic jets imaged optically or by Doppler color flow maps revealed that the laminar jet length/width ratio approached unity with an increased amount of valvular stenosis and higher volumetric flow rates. An increased amount of stenosis resulted in a higher driving pressure and smaller orifice diameter. Shorter jets were associated with larger mean Reynolds numbers. Laminar jet lengths were only imaged optically and by Doppler color flow maps in moderate and severe stenoses at low flow rates or cardiac outputs $\leq$3.0 liters/min (mean Reynolds numbers $\leq$3,470). At higher volumetric flow rates or increased levels of stenosis, a spray effect rather than a laminar jet was visualized. Hence, under carefully controlled in vitro circumstances, forward flow through a stenotic valve caused the formation of a laminar jet only when the degree of stenosis and volumetric flow rate were moderate. With a lesser degree of stenosis and lower volumetric flow rates, a flow profile of uniform width and length was observed. With a greater degree of stenosis and higher volumetric flow rates, a spray effect was observed. The Doppler color flow encoded structure and shape of flow distal to a stenotic valve may give indications as to the severity of the obstruction. Hence, jet shape in valvular stenosis was influenced by cardiac output (volumetric flow), orifice diameter and pressure gradient.

Proximal aliasing in stenotic jets. Higher cardiac outputs and increased severity of pulmonary stenosis resulted in a longer length of proximal aliasing (Fig. 7). Simpson et al. (21) also observed that a Doppler color flow acceleration zone or zone of proximal aliasing was present in a pulsatile flow model of serial subvalvular and valvular stenosis; this zone increased in length at higher flow rates. This acceleration zone corresponded to our proximal aliasing and the two behaved similarly. By knowing the length of the proximal aliasing in our model, one could not, however, predict the severity of pulmonary stenosis because there was overlap among the various valves. If the volumetric flow rate or mean Reynolds number was known in addition to the length of proximal aliasing, difficulty was still encountered in predicting the severity of valvular stenosis because the curves were close.

Proximal aliasing similar to that produced by a stenotic valve has been observed in patients with coarctation of the aorta (22). With coarctation, the proximal jet narrowing correlated with the severity of the obstruction. A narrower jet was associated with a tighter coarctation. There was no correlation between the length of proximal aliasing and severity of the coarctation, corresponding to our observation in pulmonary stenosis.

Clinical applications. This study showed that the structure and shape of stenotic and unbounded (regurgitant) jets was influenced by the orifice diameter, pressure gradient, volumetric flow, fluid viscosity and presence or absence of radial flow velocities that interact with the jet. Radial velocities that interact with jets clinically are pulmonary or systemic inflow in the instance of atrioventricular valve insufficiency and atrioventricular valve inflow that may interact with semilunar valve insufficiency. In mixed valve disease (such as aortic insufficiency and mitral stenosis) many jets can interact with one another, thereby further complicating the jet formation. Quantifying the amount of valvular regurgitation by Doppler color flow mapping alone may be difficult owing to the many factors that influence the jet; however, the profile across a stenotic valve may be easier to interpret. The severity of obstruction may be indirectly assessed by observing the Doppler color flow map distal to the valve. Mild stenosis created a minimal disturbance in the color flow profile, whereas moderate stenosis created a laminar jet and severe stenosis created a jet spray. The length of the proximal aliasing was not helpful in assessing the severity of the obstruction. Which factor or factors have the greatest influence on Doppler color jet structure and shape is difficult to determine.

Although not the purpose of this study, other factors may influence the size of Doppler color flow imaged jets and make quantitation difficult. These variables are controlled by the physician or echocardiographic technician and include the choice of transducer frequency, size of the color sector interrogated, frame rate, pulse repetition frequency, Nyquist limit of the transducer and color gain. Until these factors are universally agreed on, quantitation of Doppler color flow areas or size may vary among institutions. Hence, the quantitation of clinical valvular regurgitation or stenosis based solely on the Doppler color jet size or flow area remains problematic. The visualization of jets by Doppler color flow mapping studies can be used for a semiquantitative approach to cardiology. Additional insights into fluid hydrodynamics and instrumentation factors for flow imaging will be required if a finer degree in quantitation is to be achieved.

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