A Method for Determining the Reference Effective Flow Areas for Mechanical Heart Valve Prostheses

In Vitro Validation Studies

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Background—The anatomic opening area (AOA) is usually reported as the primary index of mechanical heart valve function. Because flow contracts immediately distal to an orifice as a result of the vena contracta effect, AOA may not be a good measure of true effective flow area.

Methods and Results—Laser flow imaging was used to visualize the contraction in the jet flow stream as it passed through bileaflet mechanical valves under steady and pulsatile conditions. Such visualization allowed clear measurement of the individual vena contracta areas (VCAs) of the 3 valve orifices. VCAs for side orifices were larger (94 ± 2% of AOA) than those through the central orifice (34 ± 8%). Formation of large radial vortices around the leaflet tips constricted the central orifice flow stream and appeared to be the main reason for smaller central VCA. Total VCA remained constant until ~0.5 orifice diameters (~1.0 cm) downstream, beyond which cross-sectional area increased as a result of entrainment of receiving chamber flow. Total VCA was larger for steady flow (89.6 ± 2.7% of AOA) than for pulsatile flow (76.3 ± 5.0% of AOA).

Conclusions—This study further clarifies flow dynamics through bileaflet mechanical valves and provides previously unavailable reference information on VCAs for these valves. Such information should aid clinicians in explaining Doppler-derived and catheter-measured pressure discrepancies, validating clinical techniques for quantifying effective flow areas, and optimizing valve size for implantation. The method should also be useful for comparative studies of different valve designs. (Circulation. 2000;101:1953-1959.)

Key Words: blood flow ▪ valves ▪ echocardiography

The noninvasive evaluation of forward flow areas for prosthetic heart valves remains an important goal for cardiologists because such information is directly related to valve function. Echocardiographic determination of anatomic orifice area may not represent the available flow area through the valve because of the vena contracta effect, which is a constriction in the flow stream and flow area as it passes through an orifice. In this regard, it is the effective orifice area, ie, the cross-sectional area of the flow at the vena contracta, that truly reflects the area available for flow. There is much confusion regarding how best to measure vena contracta area, whether vena contracta area changes with flow, and the exact definition of vena contracta area.

Accurate information on the vena contracta areas for bileaflet mechanical valves has not been available to date. A clear understanding of vena contracta dynamics and information on true vena contracta areas would help clarify discrepancies between catheter-measured and Doppler-derived pressure drops through these valves,1,2 refine existing clinical techniques such as the Gorlin and Gorlin formula and the Doppler continuity equation for effective flow area measurement,3-5 provide reference data for validating new flow quantification techniques such as color Doppler and MRI, aid in optimizing prosthetic valve size during valve replacement surgery, and establish an important clinical database for comparison with clinically measured areas, similar to the tables available for normal pressure drops for various valve sizes and models.6

The purpose of this study, therefore, was to elucidate vena contracta dynamics and areas for bileaflet mechanical valves with precise laser flow visualization (LFV) techniques. There were 4 aims for this study: (1) to validate the LFV method for measurement of vena contracta areas (effective flow areas), (2) to visualize the flow dynamics distal to bileaflet mechanical valves, (3) to determine how far downstream the vena contracta extends for such valves, and (4) to examine the effect of flow rate on vena contracta area.

Methods

Experimental Setup

Figure 1 illustrates the experimental setup. A dual-chambered transparent acrylic in vitro model was used for these studies. Flow media used was a mixture of deionized water and 30% (by volume)
glycerin mixed with neutrally buoyant silver-coated glass microparticles (mean diameter, 8 μm; fluid viscosity, 3.5 cp; density, 1.04 g/mL). Steady (30 to 100 mL/s in increments of 8 mL/s) and pulsatile (60 bpm; 40 to 80 mL/beat) flows were directed through the model. Bileaflet mechanical valves (St Jude Medical Inc) were placed in the central plate of the model. Valves had nominal diameters of 25, 29, and 33 mm (cross-sectional areas, 2.45, 3.03, and 4.06 cm²).

Instantaneous flow rates were measured with a calibrated ultrasonic flowmeter (model T109, Transonics Inc). Stroke volumes were obtained by digital integration of instantaneous flows. Proximal and distal chamber pressures were measured with solid-state pressure transducers (2.5F, model SPR 524, Millar Instruments Inc). Pressure and flow information was digitized with a data acquisition system (model NB MIO 16X, National Instruments Inc) and displayed on a Macintosh microcomputer with LabView (National Instruments Inc) software. Mean pressure gradients between proximal and distal chambers varied from 1.5 to 4 mm Hg with the valve in the open and from 10 to 24 mm Hg in the closed position. These gradients correspond to previously published data for St Jude valves (2 and 4.90 cm²) and steady flow (25 to 100 mL/s, n=5) conditions. Because the vena contracta area represents the cross-sectional area of the “fresh” flow entering through the orifice, no entrainment flow, ie, suction of receiving chamber fluid into the jet, should be present. Therefore, measurement of the velocities across the cross section of the vena contracta should allow differentiation of the vena contracta flow region, where velocities would be high, from the background flow region, where velocities would be very low. We have used this approach to identify and measure the vena contracta for computer modeling studies of flow through stenotic valves. This method has also been used to identify the vena contracta width on color Doppler images of regurgitant flows. Because of the relatively poor lateral resolution of color Doppler, we used the high-resolution technique.

**Validation of LFV Method to Measure Vena Contracta Areas**

To ensure that the LFV method provided true vena contracta areas, we performed a validation study using 3 circular orifices with areas comparable to the areas of the bileaflet valves tested (1.78, 3.14, and 4.90 cm²) and steady flow (25 to 100 mL/s, n=5) conditions. Because the vena contracta area represents the cross-sectional area of the “fresh” flow entering through the orifice, no entrainment flow, ie, suction of receiving chamber fluid into the jet, should be present. Therefore, measurement of the velocities across the cross section of the vena contracta should allow differentiation of the vena contracta flow region, where velocities would be high, from the background flow region, where velocities would be very low. We have used this approach to identify and measure the vena contracta for computer modeling studies of flow through stenotic valves. This method has also been used to identify the vena contracta width on color Doppler images of regurgitant flows. Because of the relatively poor lateral resolution of color Doppler, we used the high-resolution technique.
of digital particle image velocimetry (DPIV) to obtain cross-sectional velocity profiles of the jet immediately distal to the orifice. DPIV is now a well-established, highly robust technique to measure whole-field velocity vectors, with excellent spatial resolution (<1 mm), high velocity range (0.01 to 10 m/s), and quantitative output of results. Because of the limited scope of this paper, the reader is referred elsewhere for details on DPIV. Validation study data are given in the Results section.

Statistical Analysis
All values given are mean±SD. Linear regression of jet core area versus downstream distance was performed to examine whether jet core area changed significantly with downstream distance. Vena contracta area was also analyzed against flow rate (stroke volume for pulsatile flow conditions) through linear regression. To facilitate analysis among valve sizes, the contraction coefficient, defined as vena contracta area divided by manufacturer’s anatomic area, was also calculated. Contraction coefficients for different valve sizes were compared by use of 1-way ANOVA. Intraobserver variability was calculated by comparing the differences in 3 consecutive measurements of vena contracta area performed by the main observer. A subset of the measurements (25%) was repeated by a second observer, blinded to the previous measurements, to calculate interobserver variability. The level of significance for all statistical tests was set at 5%.

Results
Validation of LFV Method: Circular Orifices
Figure 3 displays the results from the DPIV validation studies, showing a typical DPIV image of the vena contracta...
region (Figure 3A) and linear regression between DPIV- and LFV-measured vena contracta areas (Figure 3B). Velocity magnitude in Figure 3A is represented by both vector length and vector color. The reduction in flow area as it passes through the orifice can be clearly seen, and the vena contracta width can be easily identified. It is only after ~1 orifice diameter (15 mm here) that we see entrainment flow (shown in dark blue) entering the jet. This suggests that vena contracta area can be accurately measured until 1 diameter downstream of the orifice. Excellent agreement between the 2 methods was also found (Figure 3B). For each orifice size, the close clustering of data points indicates that flow rate does not significantly affect DPIV- or LFV-measured areas. Mean percent difference between the 2 methods was 0.34±1.9%.

Mean contraction coefficients (vena contracta area/anatomic area) for these orifices ranged from 0.81 to 0.79 and agree with previously published engineering standards for orifice flow.11

**Figure 4.** Dynamic images of vena contracta area over pulsatile cycle for 25-mm valve (60 bpm; stroke volume=70 mL/beat). At cycle initiation, side orifices fill first, followed by central orifice flow. Swirling vortex structures can be seen extending into central orifice flow region. Flow area through side orifices is more stable, ie, it retains its initial structure over majority of pulsatile, than flow through the central orifice.

**Measurement of Distal Jet Core Areas: Bileaflet Valves**

Figure 4 displays the dynamic change in vena contracta area over the pulsatile cycle for the 29-mm valve obtained with the LFV technique. Time increments are shown on each image. The flow is directed out of the page. The 3 individual jet core areas and vortex structures at the top and bottom edges of side and central flow areas can be clearly seen. At the beginning of the pulsatile cycle, flow is initially directed through the side orifices (Figure 4B). The central orifice begins to fill up ~66 ms into the cycle (Figure 4C), after which side and central orifice vena contracta areas remain constant (Figure 4D through 4F) until the leaflets begin to close (t=528 to 594 ms; Figure 4G through 4I). After the initial filling and for most of the pulsatile cycle, the side orifices fill to near-maximal capacity, while the vortex structures emanating from the leaflets substantially constrict flow area through the
central orifice (see frames from 99 to 330 ms, Figure 4D through 4F).

Three separate jet cores could be visualized until ≈1.8 cm downstream of the orifice, after which the 3 jets coalesced into 1. Total vena contracta area remained constant until ≈0.5 valve diameters downstream of the leaflet tips (Figure 5). This corresponded to ≈1 cm downstream of the leaflet tips for all valve sizes tested. Linear regression analysis revealed no significant change in the minimum jet core area up to 1 cm downstream over all flow and orifice conditions (P = NS). Beyond this point, flow within the receiving chamber began mixing with flow in the jet core by the process of entrainment. This caused the cross-sectional area of the jet to increase and an obliteration of orifice shape. These findings are similar to DPIV results seen for the simple circular orifice studies, except that for the circular orifices, jet core area remains constant for a greater distance (1 orifice diameter, 1.5 to 2.5 cm) downstream. The difference is presumably due to the complex 3-dimensional (3D) nature of the bileaflet mechanical valve orifices.

No significant change in vena contracta area with steady flow rate was observed (Figure 6A). Figure 6B shows instantaneous vena contracta area measured over the pulsatile cycle for the 25-mm valve at all stroke volumes. Similar results were seen for the other valve sizes. Very little change in vena contracta area was observed over most of the pulsatile cycle. Figure 6C shows mean contraction coefficient (averaged over the pulsatile cycle) as a function of stroke volume. As for the steady flow conditions, vena contracta areas did not vary significantly with flow rate. Mean vena contracta areas for pulsatile flow were smaller (mean Cc = 0.763 ± 0.05) than for steady flow (mean Cc = 0.896 ± 0.027, P < 0.0001).

Interobserver and intraobserver variabilities for vena contracta area measurements were 3.8% and 4.2% for steady flow and 5.7% and 6.3% for pulsatile conditions, respectively.

**Discussion**

In this investigation, we first validated the use of a LFV technique to measure actual vena contracta areas and then used this method to visualize and measure vena contracta areas for bileaflet mechanical valves. The vena contracta area represents a contraction in the edges of the flow streamlines as they move through the orifice. For orifices without a smoothly tapering proximal geometry, inertia prevents proximal streamlines entering the orifice from the side from changing direction instantly. As the flow passes through the orifice, the streamlines change direction to run parallel to the...
main flow direction but not before “squeezing” the main flow and causing a constriction (vena contracta) in the cross-sectional area of flow immediately distal to the orifice. The vena contracta area thus represents the most direct measure of the flow area for a particular orifice or valve.

Many attempts have been made to use the color Doppler image of the jet region immediately distal to the orifice (proximal jet region) as the clinical measure of vena contracta width for regurgitant and stenotic lesions. One problem with the use of color Doppler is that the poor lateral resolution may not accurately delineate the region separating vena contracta and receiving chamber flow. Our DPIV validation technique provides significantly higher spatial resolution and clearly separates vena contracta flow from background flow (Figure 3). The DPIV technique could not be used to determine vena contracta areas for the bileaflet valves because of the complex flow field, which would require the highly labor- and time-intensive task of full 3D reconstruction of DPIV velocities.

The validation studies show that the LFV method measures vena contracta areas accurately and contains no flow dependence. Additionally, the DPIV results reveal that the vena contracta region remains relatively constant until ≈1 orifice diameter downstream. LFV was then performed on the bileaflet valves. Several observations were noted. First, for pulsatile conditions, flow entered through the side orifices before the central orifice. Second, side orifices fill to near-maximal capacity for steady and pulsatile flows, whereas central orifice flow is constricted severely because of the presence of vortex structures. Third, total vena contracta area remained constant until ≈0.5 valve diameters downstream of the orifice for all conditions. Fourth, there was no significant change found in vena contracta area with flow rate for steady or pulsatile flow state. Finally, steady-flow vena contracta areas were higher than mean pulsatile flow areas.

The dynamics of the vena contracta region have been studied in the design of orifice flow meters, regurgitant and stenotic cardiac lesions, and various jet flow investigations. Fluid mechanics predicts that the vena contracta region for an orifice should remain constant until a short distance downstream; this is called the jet core region. Within this region, the viscous shear layer developed at the jet boundary has not penetrated into the jet core. The result is that the shape and size of the jet core mimic orifice shape and size. Our LFV method takes advantage of this fact by visualizing the flow cross section at the jet core region. The length of the vena contracta region for bileaflet mechanical valves is even smaller (<0.5 diameter) than that for planar orifices (<1 diameter). This could be due to the 3D nature of the valve orifice, which allows the initial development of the shear layer within the valve rather than immediately downstream. This does not present a practical problem, however, because for most valve sizes (>19 mm), the region for vena contracta measurements should extend to 1 cm.

The question of whether vena contracta area varies with flow rate remains controversial. Investigators have found variations in Doppler- and Gorlin-calculated valve areas in vitro and clinical studies. Our results for these prosthetic valves reveal no significant variation in vena contracta area with mean flow rate or stroke volume. We have observed a similar lack of flow dependence in previous studies using computer modeling and in vitro experimentation of flow through valvular stenosis. This was also confirmed by the DPIV data. Our results provide evidence that variations in Doppler or Gorlin areas with flow may be a consequence of methodology limitation rather than true vena contracta variation.

It has been known that flow dynamics and downstream velocities are different for the side versus central orifices for the bileaflet mechanical valve. However, no direct visualization of the actual vena contracta flow area through the valves has been reported. We observed stable flow through the side orifices but disturbed flow through the central orifice. Central flow disturbances are presumably due to the 3D expansion-type shape of the central orifice, which, like a flow diffuser, causes an increase in pressure and the potential for adverse pressure gradients, flow separation, and vortex formation to occur. In fact, we observed severe vortex-type disturbances within central orifice flow. The side orifices are primarily responsible for transporting most of the flow through the valve.

**Study Limitations**

There are several limitations that should be acknowledged in this study. This is an in vitro study, and as with any experimental study, it was not possible to reproduce all physiological variables. For example, we did not simulate the complex physiological loading of the valve in open and closed states. The valve was mounted in a rigid position, and the working fluid simulated blood under high shear (constant viscosity). The pulsatile flow conditions may not absolutely mimic physiological pressure and flow waveforms.

**Clinical Implications**

The question of how best to evaluate mechanical valve function remains controversial. Although measurement of transvalvular pressure gradients with Doppler remains the method of choice, there is still confusion as to the meaning behind this parameter. For example, investigators have shown that the pressure gradient differs between side and central orifices for the bileaflet mechanical valve. Maximal pressure drop and hence maximum velocity have been shown to occur through the central orifice; however, sampling this gradient may produce a false estimate of overall pressure drop through the valve because the central orifice contains the smallest vena contracta area. Our results point to the side orifices as the most important conduit for flow and indicate that pressure gradients should be measured through these orifices. This is echoed in a clinical study by Vandervoort et al., who suggested sampling the side orifice pressure gradient as the most reliable reflector of overall transvalvular dynamics. The measurement of gradients, however, can be problematic in the presence of changing preload and/or afterload in which the pressure gradient may vary with no alteration in valve function. Therefore, we believe that effective flow area, if accurately measured, provides the most accurate means of clinically evaluating valve function. Recently, Leung et al. used proximal flow convergence methods to estimate effective orifice area for bileaflet mechanical
valves in patients. These authors point to the lack of a universal gold standard for comparing clinically measured effective orifice areas as a continuing limitation for clinical studies, especially because results from Gorlin, pressure half-time, and continuity may vary widely for the same valve. The vena contracta results presented here can be used as reference data for comparing effective orifice areas measured with existing clinical methods, especially because we have shown that actual vena contracta areas exhibit no flow dependence over a wide range of flow rates. Finally, our results should lay groundwork for the development of new clinical imaging techniques for vena contracta visualization and measurement. We have shown using in vitro studies that 3D echocardiography coupled with contrast provides a novel method is a direct parallel of our LFV technique in that the ultrasound “sheet” is projected at the short axis of flow (perpendicular cross section) immediately distal to the valve and that the contrast agent acts as the “reflective” particles that will be imaged echocardiographically at the instant they cross the viewing plane. Such echo-based flow visualization methods should allow direct visualization and quantification of the vena contracta in the clinical situation.

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References
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