THE EFFECT OF VARYING DEGREES OF STENOSIS ON THE CHARACTERISTICS OF TURBULENT PULSATILE FLOW THROUGH HEART VALVES

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Abstract — Many problems and complications associated with heart valves are related to the dynamic behavior of the valve and the resultant unsteady flow patterns. An accurate depiction of the spatial and temporal velocity and rms distributions imparts better understanding of flow related valve complications, and may be used as a guideline in valve design. While the generalized correlation between increased turbulence level and the severity of the stenosis is well established, few studies addressed the issue of the intermittent nature of turbulence and its timing in the cardiac cycle, and almost none assessed the effect of a progressive stenosis on the flow characteristics through heart valves. In this experimental work we simulated the type of flow which is present in normal and stenosed valves and conducted a comprehensive investigation of valve hemodynamics, valvular turbulence and morphology under varying degrees of stenosis. The characteristics of valves and stenoses were simulated closely, to achieve the flow conditions that initiate turbulent flow conditions. Laser Doppler anemometry (LDA) measurements were carried out in a pulse duplicator system distal to trileaflet polyurethane prosthetic heart valves, installed at mitral and aortic positions. The effect of the degree of the stenosis was comparatively studied through the structure of the turbulent jets emerging from normal and stenotic heart valves. Maximum turbulence level was achieved during the decelerating phase and correlated to the severity of the stenosis, followed by relaminarization of the flow during the acceleration phase. The intermittent nature of the turbulence emphasized the importance of realizing the timing of the turbulence production and its spatial location for optimizing current valve designs. The plug flow through the normal aortic valve prosthesis was replaced by jet like behavior for a 65% stenosis, with the jet becoming narrower and stronger for a 90% stenosis. The morphology of the velocity and turbulence waveforms was found to be governed by the stenosis geometry and the valve position (aortic, mitral).

Keywords: Stenosis; Bloodflow; Heart valves.

INTRODUCTION

The presence of turbulence in the cardiovascular system is generally an indication of some type of abnormality. Most cardiologists agree that turbulence near a valve indicates either valvular stenosis or regurgitation, depending on the phase of its occurrence during the cardiac cycle. In the western society, the commonest and most clinically important form of heart valve stenoses are in the aortic valve. Late in the disease they can cause progressive damage to the heart muscle and result in a heart failure. Flow characteristics through heart valves such as localized fluctuations, turbulence intensity, distorted flow patterns and large local velocity gradients, influence factors that may contribute to the etiology of cardiovascular disease. As stenosis progresses, its effect is apparent through the changing morphology of the flow characteristics. Therefore, a thorough hemodynamic characterization and evaluation of the effect of varying degrees of stenosis is essential.

The nature of the flow through the aortic valve is such that the valve's leaflets vibrate during diastole (towards closure), especially in stenosed valves where the flow pattern tends to be of turbulent nature. Although semi-active (by its papillary muscles), the mitral valve's two leaflets are also subjected to such kind of vibrations during diastole (Padula et al., 1968). Turbulence was observed in human subjects in the region of a normal aortic valve during ejection, in the presence of a high cardiac output. Distal to a stenotic valve turbulence extends throughout the major portion of the ejection (Sabbah and Stein, 1976).

Fluid mechanics of complex cardiovascular flow situations are extremely difficult to obtain experimentally in vivo, and difficult to simulate in vitro. Nevertheless, an in vitro experimental set-up enables a comparative evaluation under controlled conditions. Numerous in vitro measurements were conducted in pulse duplicator systems in order to investigate the flow field through prosthetic heart valves. The quality of the simulation vary according to the investigators and the pulse duplicator systems involved (Chandran et al., 1983, 1989; Chen et al., 1988; Woo et al., 1983a; Yoganathan et al., 1979a, 1983).

Steady-flow analyses of in vitro testing should be treated with particular caution. Comparative study of pulsatile flow and steady flow, showed that the rms level measured in steady flow tended to be an upper
bound on that measured for pulsatile flow (Phillips et al., 1980; Woo et al., 1983a, 1986; Yoganathan et al., 1983, Chen et al., 1988; Hanle et al., 1989). This finding indicates that significant portions of the pulsatile-flow cycle generally cannot be inferred from steady-flow studies. While steady-flow measurements assume highest rms level and shear stress at peak flow, it appears that the magnitude of the fluctuating velocities is dependent on the systolic phase. Late in the systole (deceleration phase) the flow was highly disturbed with high rms levels that would have been impossible to predict a priori (Hanle et al., 1986).

Measurements distal to a polyurethane trileaflet prosthetic heart valve mounted in the mitral position showed that during systole the trileaflet valve produces a strong jet towards the apex (Schoephoerster and Chandran, 1991).

Previous in vitro models have used nozzles and other simulations of stenosis, without replicating the in vivo dynamic characteristics of aortic stenosis in which energy is lost at the inlet of the stenosis, because of the energy required to open thickened leaflets (Gorlin, 1987) and the turbulence created by eccentric orifice areas (Flachskampf et al., 1990). Both Schoephoerster et al. (1989) and Sprigings et al. (1990) simulated severing stenosis by progressively suturing the valve leaflets. The former have shown that the original Gorlin formula underestimates the stenosis area by not taking into account the changing dynamics of the valve opening and closing behavior, the stiffening of the leaflets, and dependence of the valve closure on the viscous effects of the jet flow created by the stenosis. The latter correlated the left ventricular stroke work loss to the severity of the stenosis.

Many problems and complications associated with heart valves are related to the dynamic behavior of the valve and the resultant unsteady flow patterns. Stenotic flow conditions in heart valves are manifested through the changes in the morphology of its flow characteristics. While the correlation between the severity of the stenosis and elevated turbulence levels is a well-established fact, few of the researchers addressed the issue of the changing flow patterns during the pulsatile cycle, and almost none conducted comparative measurements in order to assess the effect of a progressive stenosis on the flow characteristics. A depiction of the changing flow dynamics and the turbulent characteristics of the valvar waveforms distal to progressively stenosed valves imparts better understanding of flow related valve complications associated with severing stenosis. The recent study is aimed at studying and comparing the flow field characteristics distal to progressively stenosed trileaflet polyurethane valves installed at the aortic and mitral positions in a pulse duplicator system. The changing dynamic properties of the different stenoses were carefully mimicked, so that the geometric and the dynamic effects of the stenosis on the morphology of the flow field could be realized.

**EXPERIMENTAL METHODS**

Velocity measurements were made using a LDA system employed in the dual beam, forward scatter mode. The system included the following components (TSI, USA): Argon-Ion 2 Watt laser-continuous 514.5 and 488 nm wavelengths, optic package and a counter signal processor. At the intersection of the beams an ellipsoidal sampling volume with dimensions of 0.75 mm x 0.072 mm and 2.67 μm fringe spacing, in air, was produced. Polystyrene latex spherical beads with average diameter of 3.134 μm (LB-30, SIGMA Chemical Company, U.S.A.) were used to seed the flow field.

Measurements were conducted in a modified Caltech pulse duplicator-type system. Description of the Caltech pulse duplicator appear elsewhere (Yoganathan et al., 1991). The modification was aimed to achieve a better simulation of the physiological flow conditions through heart valves. The left ventricle compartment (Fig. 1) was designed to facilitate LDA measurements inside the ventricle distal to the mitral valve, as well as to simulate the contraction features of the left ventricle being the driving force of pulsatile blood flow in the left heart circulation. A pulsatile air pump (Harvard Apparatus Dual Phase Control model 55-3305 pulsatile blood pump, modified to be used with air instead of liquid) generated volume time curves that drove the left ventricle’s bladder. Under characteristic operation conditions the pulse duplicator system produced a quasi-physiological left ventricular and aortic pressure waveforms (Fig. 2), with a volumetric discharge of 5.81 min⁻¹. As our interest for aortic valve measurements lay in the ejection characteristics, diastole was chosen to be short compared to systole (Fig. 2(a)). Similarly, for mitral valve measurements, systole was chosen to be short compared to diastole (Fig. 2(b)). The left ventricle compartment was installed on a computer controlled, three coordinates stepper motor mechanism, and was displaced during measurements relative to the fixed LDA measuring volume.

Measurements distal to the aortic valve were carried out along the symmetry axis of an 80 mm long rigid lucite conduit (16 mm ID), connected to the valve stent (Fig. 3). Three sinuses were carved in the valve’s lucite seat, enabling the installation of a trileaflet prosthetic heart valve, complete with its molded sinuses (very closely simulating the geometry of the three sinuses of Valsalva in the aortic root) manufactured from polyurethane by Abiomed company, U.S.A., according to the specifications (Fig. 3) (details on this valve can be found in Woo et al., 1983b). Their study showed that the flow characteristics of synthetic valves were similar to the bovine pericardial bioprostheses. Measurements in the left ventricle, distal to the mitral valve, were carried out in the rigid half of the compartment along its symmetry axis, thus avoiding optical distortions.
Effect of stenosis on turbulent pulsatile flow through heart valves

Simulation of aortic valve stenosis was achieved by a partial coating and adhesion of the valve's leaflets convex side with flexible silicon rubber. In this manner a physiological-like stenosis geometry and dynamic behavior (mass and inertia of the leaflet) was achieved, coupled with the morphological changes related to tissue overgrowth caused by fibrosis and thrombus formation which are common occurrences in valvular stenosis. Mitral stenosis was simulated by a partial suturing of the valve leaflets. Combined simulation of a severe aortic stenosis with heavily calcified leaflets and incompetent valve behavior was achieved by coating the leaflets with Araldite epoxy adhesive, thus creating stiffened leaflets configuration with a fixed 90% stenosis. This configuration was comparatively studied with other stenosis configurations in order to single out the effect of the flexibility of the leaflets in generating flow fluctuation.

Phase locked ensemble data averaging was performed during successive cycles of the pulsatile flow.
A pressure transducer (Data Instruments model SA) was connected to the left ventricle. The signal amplitude and slope were processed for the selection of time window duration. With time window defined, this box activated an internal reference clock in the computer/counter LDA interface card (Zech Electronics model 1400), which was interpreted through software as phase angle information. The triggering pulse inhibited LDA data acquisition when time window was closed, and enabled simultaneous acquisition of LDA data and phase angle information when open.

Instantaneous phase locked point velocities were measured on-line and were accumulated in segments which constituted the time window. The segments were software chosen so that 360° corresponded to 1 s. The time resolution for segment period was determined by the LDA interface card internal timing reference which was chosen as 156,250 Hz, leading to segment period time resolution of 6.4 ns. Up to 50 successive cycles were sampled in a given spatial location in the flow for the segmental phase locked ensemble averaging. Output included mean velocity and rms (turbulence intensity) waveforms. Typically, the sequential waveforms were measured at 25 radial locations for each cross section, and grouped together to create a three-dimensional surface depicting the radial and temporal distribution of the phase averaged axial velocity and rms. A similar presentation was used by Reul et al. (1986) in order to depict the dynamics of the flow patterns (the three-dimensional surfaces were called 'tidal development of downstream velocity profiles during systole' in their terminology). The three-dimensional perspective is shown either as one cycle between two successive peak systoles, or as one cycle between two successive diastoles, depending on whether the data on the aortic or mitral valve is presented. In this presentation, the rms of each phase was normalized by the corresponding local mean phase velocity, so that its dynamic changing pattern would be more pronounced. A typical flow cycle was achieved with a pulsatile pump rate of 75 BPM (beats per minute), which corresponded to a flow cycle of 800 ms. The processed pressure transducer signal yielded a time window of 700 ms duration, so that only a minor part of the cycle was excluded. The time in the cycle was presented in terms of phase angle, where 360° corresponded to 1 full second (156,250 beats of the LDA interface internal clock) so that the time window was presented by a typical phase angle value of 252°.
RESULTS

The unsteady LDA velocity data of the flow field distal to a normal aortic valve depict the dynamics of normal valvular flow characteristics during the cycle (Fig. 4), measured at a cross section 25 mm distal to the valve. This was the closest site possible for LDA measurement in the proximity of the aortic valve, without interference with the three Valsalva sinuses. $Re$ (based on peak jet velocity through the valve $- 1.36 \text{ m s}^{-1}$, and the valve's hydraulic diameter computed from the maximal orifice area $- 13.8 \text{ mm}$) was $18,818$. The cross section of the three-dimensional perspective of the velocity waveform shows the radial distribution of the velocity across the aorta, measured in its axisymmetric plane. It reveals the radial velocity profile during peak systole ($\sim 0^\circ$), which resembles that of a jet-like plug flow. The effect of the valve geometry on the velocity distribution in the proximity of the valve is apparent, with leaflets-induced velocity spikes clearly visible. The characteristic steep radial velocity gradient appears proximal to the aortic wall. The axial (temporal) velocity drops sharply during the decelerating phase (around $140^\circ$). The following acceleration phase is more moderate, with velocity rising in a less steep manner.

In the three-dimensional rms perspective (Fig. 4), the rms level elevates toward the deceleration phase, peaks following the rapid closure of the valve (around $150^\circ$) and maintains a high level until the flow acceleration build up commences (between phase angles $175-200^\circ$), leading to a decrease in the rms level towards relaminarization of the flow. One can see clearly the turbulence peaks in the shear layer zone ($2r/d = 1$, where $r$-characteristic radius of the fully open valve orifice is $6.9 \text{ mm}$). The radial positions for the peaks are about 3.5 and 11.5 mm from left wall, corresponding to the locations of the steep radial velocity gradient. In the centerline zone ($2r/d = 0$) the rms level during systole ($0-100^\circ$) is the lowest, resembling a jet potential core behavior.

The stenosis has a noticeable effect on the morphology of the velocity and rms waveforms, with elevated turbulence levels and a distinct jet-like behavior, as revealed by the cross section of the three-dimensional perspective of the velocity and rms distributions measured $30 \text{ mm}$ distal to an aortic valve with $65\%$ stenosis (Fig. 5). The stenosed leaflets are less flexible and have more inertia due to the silicon rubber coating, therefore their effect on the cross-sectional velocity profile is less pronounced. The leaflets do not deflect to their fully open position, creating a smaller orifice with an hydraulic diameter of $11.14 \text{ mm}$ when fully open. Consequently, the jet-like behavior is more apparent, with peak velocities exceeding $1.5 \text{ m s}^{-1}$. The velocity gradient toward the wall is less steep. The cross-section of the three-dimensional rms perspective during peak systole exhibits the shear layer rms peaks, indicating that the shear layer extends now towards the core region. The potential core of the jet is characterized by a plateau in the rms centerline zone, extending till the beginning of the decelerating phase (around $100^\circ$, $30 \text{ mm}$ distal to the valve). Downstream there is a small lag in the initiation of the deceleration phase (around $135^\circ$, $50 \text{ mm}$ distal to the valve).

Velocity and rms distribution $25 \text{ mm}$ distal to an incompetent valve with $90\%$ stiff stenosis (Fig. 6, shown from diastole to diastole) exhibit characteristics of turbulent jet produced by the severe stenosis. The effect of the oddly shaped stenosis geometry is clearly seen in the velocity profile. During peak systole a characteristic turbulent jet flow velocity profile is observed. The jet width is significantly narrower than the one created by the $65\%$ stenosed configuration, a fact that can be attributed to the much smaller hydraulic diameter ($4.37 \text{ mm}$). The overall rms level is high throughout the flow cycle with almost no potential core associated rms plateau in the centerline zone, indicating that the jet is fully turbulent. The steep radial velocity gradient towards the wall, characteristic of a fully turbulent jet velocity profile, creates shear layer rms peaks which extend throughout the systole.
Fig. 5. Perspective of radial and temporal velocity and rms distributions 30 mm distal to an aortic valve with 65% stenosis (shown between two successive peak systoles). The jet-like behavior is more apparent. The overall rms level is elevated and the shear layers rms peaks are more pronounced.

A normal prosthesis in the mitral position produces during systole a jet towards the apex (Fig. 7, shown from diastole to diastole). At 15 mm distal to the valve, the effect of the valve geometry on the cross-sectional radial velocity distribution is apparent during the acceleration phase. During systole (60–190°), a jet-like velocity profile can be identified. The rms distribution is asymmetrical. In the proximity of the ventricle wall a sharp shear layer peak is maintained throughout the flow cycle (radial coordinate—15 mm). At the interface between the unbounded side of the jet (along the opposite side of the wall, radial position—2.5 mm) the shear layer rms peak is less pronounced. A potential core with its characteristic rms level plateau can be seen along the centerline, disappearing during the deceleration phase. The 65% mitral stenosis (Fig. 8, shown between two successive diastoles) accelerates the flow, creating a strong jet flow during systole. The stenosis produces elevated rms levels during the deceleration phase (90–135°). A delay between the deceleration and acceleration phase is noticeable.

DISCUSSION

Many problems and complications associated with heart valves are related to the dynamic behavior of the valve and the resultant unsteady flow patterns. Three-dimensional visualization of the velocity and rms distributions is considered rendering more comprehensive information about the flow field and its dynamics. A more complete picture of the hemodynamics of heart valves imparts better understanding of flow related valve complications, and may be used as a guideline in valve design. Stenotic flow conditions in heart valves are manifested through the changes in the morphology of its flow characteristics. While the correlation between the severity of the stenosis and elevated turbulence levels is a well-established fact, few of the researchers addressed the issue of the changing flow patterns during the pulsatile cycle, and almost none conducted comparative measurements in order to assess the effect of a progressive stenosis on the flow characteristics. In this work comparative studies of the flow dynamics past polyurethane valves with different degrees of stenosis were conducted. The investigation focused on the effects of varying degrees of stenosis on the velocity and turbulence waveforms, and their evolution distal to the valves. Stenosis was simulated on valves installed at aortic and mitral positions by progressively gluing the leaflets together to create a smaller orifice. Detailed spatial and temporal velocity and rms distributions were measured distal to the valves to create the three-di-
Fig. 7. Perspective of spatial and temporal velocity and rms distributions 15 mm distal to a normal mitral valves (shown during diastole). The effect of the ventricle wall is noticeable through the apparent rms radial distribution asymmetry.

Fig. 8. Perspective of spatial and temporal velocity and rms distributions 15 mm distal to a mitral valve with 65% stenosis (diastole to diastole). Notice the delay in the valve reopening. The jet-like behavior is more apparent and the rms levels are higher compared to the normal mitral valve.

dimensional surfaces in which the morphology of the flow characteristics was examined. The comparative studies enabled us to examine the effect of the stenosis progression on the flow patterns produced by the valves. For both normal and stenosed valves turbulence peaked during the deceleration phase. During the acceleration-phase flow disturbances dissipated and the flow relaminarized. Stenosed valves exhibited elevated turbulence levels that correlated to the severity of the stenosis, and characteristic jet flow conditions. The morphology of the velocity waveforms reflected the reduced flow rates through the stenoses.

The normal polyurethane aortic valve generates centralized flow with a relatively flat profile similar to the natural aortic valve. The general morphology of the three dimensional velocity surfaces and the peak velocities were similar to those measured by Reul et al. (1986) distal to Ionescu–Shiley bioprosthetic and Hancock porcine valves. The shape and orientation of the spatial velocity profile (radial distribution) depicts the effect of the constriction that is generated by the valve's leaflets on the velocity profile, in a similar manner to that measured by Hanle et al. (1986) distal to a Ionescu–Shiley bioprosthetic aortic valve. The jet-like plug flow and the corresponding measured velocities are similar to those measured by Chandran et al. (1989) with trileaflet polyurethane valves from the artifical organs program at the University of Utah.

The intermittent nature of turbulence production during the flow cycle demonstrates how pulsatility suppresses turbulence in the cardiovascular system. The instantaneous centerline velocity waveform during systole (Fig. 9(a)) reveals the decrease in rms level during acceleration, and the rms plateau during systole that characterizes the potential core of the jet produced by the orifice. As the cycle approaches the deceleration phase (early diastole), the rms level rises across the aorta and peaks just before the onset of the successive acceleration phase (systole), followed by flow relaminarization and an immediate decrease in the rms level (in agreement with the measurements of Yamaguchi et al., 1988). During the deceleration phase the instability mechanism characteristic of pulsatile flow (Winter and Nerem, 1984) is responsible for the production of larger flow fluctuations (relative to the local-phase velocity), in spite of the decrease in the local-phase Re number. This differs from the case of
Fig. 9. Effect of stenosis on centerline velocity and rms waveforms during systole (30 mm distal to aortic valves). The deceleration phase for the 65% stenosis happens earlier in the cycle resulting in reduction of the flow rate through the stenosed valve. The flow rate through the incompetent valve with 90% stenosis is significantly reduced, and rms level is increased throughout systole.

steady flow, where transition to turbulence and an increase in the rms level are directly correlated to an increase in $Re$ value. In this manner when stenotic conditions give rise to increased turbulence, its intermittent nature postpones the damage to the blood constituents.

The velocity waveforms through stenosed valves reflect the reduction in blood flow through the stenosis. Compared to a normal valve the deceleration phase of the 65% stenosed valve (Fig. 9(b)) occurs much earlier in the pulsatile cycle (approximately 200° for an aortic valve with 65% stenosis and around 240° for a normal aortic valve), while the onset of the acceleration phase occurs at a similar phase angle for both valves. This implies that the stenosed aortic valve is open for a shorter period, leading to a significant reduction of blood flow discharge through the valve. The stenosed valve requires higher pressure gradients to maintain its fully open position at peak systole. If the left ventricle is not controlled to maintain the additional pressure gradient, as was the case in our pulse duplicator system, the valve will collapse sooner into its closed position. One can also observe that while the acceleration is gradual for the normal aortic valve, it is initiated a bit later for the stenosed valve and then builds up in a steeper manner. This is
caused by the additional threshold pressure gradient needed for reopening the stenosed leaflets. It can also be noticed that for the stenosed valve there is a certain recovery of forward velocity immediately after the deceleration phase (210–240°). The rms centerline waveform of the aortic valve with 65% stenosis reveals much higher rms values during deceleration, as compared with a normal aortic valve.

For the 90% stiff stenosis (Fig. 9(c)) that shows a simulation of an incompetent valve in its most severe pathological situation the valve leaflets are not flexible anymore, and are fixed in a semi-open position, creating a small orifice. Consequently, the valve loses its dynamic properties and the velocity waveform is governed mainly by the pulsatile driving force of the pump. Although a threshold pressure is not required now to open the leaflets, additional pressure is needed to overcome the resistance offered by the obstruction created by the stiff stenosis. As a result the acceleration phase is much more gradual and the successive deceleration phase shortly follows. A very short peak systole phase is maintained, resulting in a significant reduction of flow discharge through the valve. Due to directional ambiguity of the LDA signal in the absence of a Bragg cell in our LDA system, the small peak following the deceleration phase could be attributed to regurgitant flow that further reduces the flow rate. The elevated turbulence rms level throughout the flow cycle is characteristic of the turbulent jet created by the small orifice, and is much higher than the corresponding rms levels of the normal and the 65% stenosed valves.

The trileaflet valve in the mitral position produces a jet directed towards the apex of the left ventricle, similar to that measured by Schoephoerster and Chandran (1991). Unlike the aortic valve, this jet is unbounded. Nevertheless, the effect of the ventricle wall is noticeable through the apparent asymmetry in the radial velocity and rms distributions. With 65% stenosis the rms level is elevated and there is a little evidence for the existence of a potential core rms plateau during peak systole, indicating that the jet is developing into a turbulent one. This configuration produces a delay between the deceleration and the acceleration phase. Between 90° and 150° the flow through the valve is minimal. The reopening of the valve leaflets and the resulting acceleration phase are delayed.

Stenotic conditions were characterized by the prolonged period in which the valves remained closed as compared to normal valves, indicating the significant reduction in flow discharge through stenosed valves. Velocity waveforms measured in stenotic valves reflected the reduction in the fluid dynamic efficiency of the valve’s opening and closure mechanism. The different dynamic behavior of the stenosed valves was caused by the leaflets augmented inertia, the change in the stenosis geometry and the higher pressure gradient needed to overcome the resistance to flow. It is clear that a heart with stenotic valve(s) should develop higher ventricular pressure in order to provide sufficient blood discharge to the body. In turn, highly disturbed flow conditions will arise.

REFERENCES


