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Experimental Investigation of Global and Squeeze Flow in Mechanical Heart Valve

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Abstract

The investigation of the flow field generated by mechanical heart valve (MHV) is necessary to explain the relationship between the hemodynamic performance and hemolytic and thrombogenic complications in MHVs replacement. The flow dynamics in one cardiac cycle through a transparent bi-leaflet MHV model was investigated by Particle Image Velocimetry (PIV) technique. A novel triggering method was developed to enable the simultaneous measurement of the occluder dynamics and the flow field. The shear stress distributions were also calculated at every time step of the cardiac cycle. The maximum stress was found around the hinge area at peak systolic instant, whose magnitude is larger than the threshold for hemolysis.

Squeeze flow, which occurs immediately before occluder /housing impact was particularly studied using a microscope incorporated PIV system. It was hypothesized that MHV cavitation, which is induced by the abrupt pressure reduction at the last instant of MHV closure, is closely related with squeeze flow phenomenon. Series of images within the continuously narrowing flow channel between the occluder and housing were captured. When the flow channel is reduced to around 0.1mm, flow velocity of up to 2 m/s is observed, with the strongest vorticity near the leading edge tip of the occluder. This high-speed jet flow would be accompanied by a corresponding reduction in local pressure, and hence a likely source for potential cavitation.

The influence of the geometrical profile of the valve tip for the squeeze flow was investigated with three sets of transparent bileaflet MHVs. The closure flow field was measured for each MHV model and correlated with the negative pressure transients (NPT) recording in the vicinity of occluder/housing. The results show that different valve tip designs affect the occluder closing phenomenon, including squeeze flow, occluder sudden stop and rebound. The magnitude of the NPT occurring at MHV inflow

side is also found to be related with the valve tip designs. The 60° valve tip is observed with maximum value of the first NPT peak, which may attribute to its highest squeezing flow velocity. On the other hand, the 30° valve tip has the largest magnitude of the second NPT peak, possibly causing by the more dramatic velocity change at the occluder sudden stop and rebound.

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Symbols

ρ	Density of fluid
P	Pressure
M	Coefficient of magnification
ℓ_{image}	Size of interrogation spot
V_p	Velocity of seeding particle
t_{sep}	Laser pulse separation time
d_e	Effective image diameter of particle
d_s	Diameter of diffraction spot
d_p	Diameter of seeding particle
NA	Numerical aperture of microscope lens
λ	Wavelength of laser light
$f^\#$	f number
Δx	Measurement uncertainty
$\dot{V}_p$	Time derivative of particle velocity
ρ_p	Density of seeding particle
ν	Kinematic viscosity of fluid
N_I	Image density in interrogation region
Δz	Thickness of laser sheet
u_i	Instantaneous axial velocity of flow
v_i	Instantaneous transverse velocity of flow
U	Average axial velocity of flow
V	Average transverse velocity of flow
N:	Number of cardiac cycles
τ_{xy}	Reynolds shear stress
τ_{xx}	Reynolds normal axial stress
τ_{yy}	Reynolds normal transverse stress
α	Occluder angular position
Δt	time interval

Abbreviations

AHV:	Artificial heart valve
MHV:	Mechanical heart valve
PyC:	Pyrolytic carbon
BS:	Bjork-Shiley
MH:	Medtronic Hall
SJM:	St.Jude medical
ED:	Edward-Duromedics
SB:	Sorin Biocarbon
MP:	Medtronic parallel
RBC:	Red blood cell
PIV:	Particle image velocimetry
DPIV:	Digital particle image velocimetry
LDA:	Laser doppler anemometry
FEA:	Finite element analysis
NPT:	Negative pressures transient
HFO:	High frequency oscillations
VC:	Ventricular sac
FL:	Focal length
CCD:	Charge couple device
FOV:	Field of view
NaI:	Sodium iodide
RSS:	Reynolds shear stress
Re:	Reynolds number

Chapter 1 Introduction

The replacement of the diseased heart valves with artificial heart valve (AHV) has become an effective and common treatment since the successful implantation of the first prostheses in 1960. During the past four decades, over 50 different designs have been developed and nearly three million AHV have been implanted worldwide (Yoganathan, 1996). The research work in AHV has been an active field for decades, from which basic understanding of the valve's hemodynamic performance are acquired. However, some problems remain not fully understood due to their complexity and limitations of the technology.

The present experimental study was undertaken with the aim to better understand dynamic flow characteristics across bi-leaflet mechanical heart valve (MHV). Particular attention was paid to the MHV closure dynamics in order to obtain better understanding for the relationship between cavitation and squeeze flow phenomenon.

In this chapter, some background knowledge about human circulatory system and heart structure are introduced. Following, MHV and its related fluid dynamic problems are discussed. Finally the objective and scope of this work is presented.

1.1 Circulatory System and Heart Anatomy

The circulatory system is a transport system that supplies O_2 and substances absorbed from the gastrointestinal tract to the tissues, returns CO_2 to the lungs and other products of metabolism to the kidneys. The blood circulation functions in the regulation of body temperature and distribution of agents for cell functions. The circulatory system mainly consists of systematic circulation and pulmonary circulation. For the systematic circulation, blood is pumped from the left ventricle to the arteries, arterioles and capillaries, then through venules into the veins and back to the right atrium, while for

the pulmonary circulation, blood flows from the right atrium to the right ventricle, and then is pumped through the vessels of the lungs to the left atrium and left ventricle, as shown in Fig.1.1.

Fig.1.1 Systematic and pulmonary circulations
(from Guyton & Hall, Textbook of Medical Physiology, 1996)

The heart is the pump for cyclic blood circulation. It consists of four chambers — right atrium, right ventricle, left atrium and left ventricle. The two atria and two ventricles are separated from each other by inter-atrial septum and inter-ventricular septum respectively. Blood flows through the heart in one direction: from the atria to the ventricle and out the great arteries. This one-way traffic is guided by the presence of four heart valves, namely: aortic and mitral valves, tricuspid and pulmonary valves, as in Fig.1.2. The mitral and tricuspid valves, both of which are also called atrio-ventricular valves, are at the junction of the atrium and ventricular chambers on each side. The aortic valve is at the junction of the aorta and left ventricle, while the

pulmonary valve guards the opening between the right ventricle and the pulmonary artery.

Fig.1.2 Heart structure
(from Starr & Cecie, Human Biology, 2003)

The left heart normally attracts more interests due to the higher incidence of diseases. It consists of a low-pressure collection tank known as the left atrium and a high-pressure pump known as the left ventricle. As oxygenated blood returns from the lungs through the pulmonary veins it collects in the left atrium. During the iso-volumic relaxation phase of the left ventricle, the pressure in the ventricle falls rapidly. When the pressure in the left ventricle falls below the pressure in the left atrium, the mitral valve opens, and blood flows into the ventricle. The period of relaxation and ventricular filling is known as diastole. During the filling period, blood flows continuously into the ventricle. Most of the ventricular filling during diastole occurs passively. At the beginning of the

isovolumic contraction phase of the left ventricle, the mitral valve closes, and the pressure rises in the ventricle. A properly functioning of mitral valve prevents any blood flow from the ventricle into the atrium during this period. When the pressure in the left ventricle rises above the pressure in the ascending aorta, the aortic valve opens and blood is ejected into the aorta. The period of iso-volumic contraction and ejection is known as systole. At the end of systole, the aortic valve closes and prevents any blood flowing back into the ventricle. Generally speaking, the cardiac valves are remarkably efficient devices, offering very little resistance to flow when they are open, and are capable of closing in response to a small pressure difference and with negligible regurgitant flow.

Fig.1.3 Cardiac Events
(from Guyton & Hall, Textbook of Medical Physiology, 1996)

Fig.1.3 is the illustration of typical events occurring in one cardiac cycle with a heart rate of about 78 beats/min. For a normal human being at rest, the atrium pressure is between 0~10 mmHg, while the ventricular pressure between 0~120 mmHg. And the aortic pressure between 80 and 120 mmHg. The cardiac output under these conditions is

about 4~6 L/min. The systole period is about 30~42% of the full cycle time. Whereas under vigorous exercise condition, the normal heart may have a cardiac output up to 35 L/min and a heart rate of 180 beats/min.

The normal adult mitral valve measures about 35mm across its major diameter, and the aortic valve measures about 25~27 mm. The average thickness of the valve leaflet is about 0.1mm. The mean internal diameter of the ascending aorta is about 25.4 mm and it has a wall thickness of about 2mm. At normal cardiac output, the mean pressure drop across a normally operating aortic valve is about 2 mmHg, while the corresponding pressure drop across a normal mitral valve is less than 1 mmHg.

Blood flowing inside the circulatory system is a non-Newtonian fluid, which can be assumed to be Newtonian in the arteries with a diameter larger than 10mm. The viscosity of the blood is about 3~5 cP and a density of 1.05 g/cm³.

1.2 Artificial Heart Valve

Although the natural heart valves are subject to prolonged operating condition, they normally can sustain for human's lifetime due to their excellent hemodynamic performance and self-maintaining capability. Unfortunately, some diseases, like rheumatic fever, ischemic heart disease, bacterial or fungal infection, connective-tissue disorders, trauma and malignant carcinoid, can possibly impair their functions. It leads to various disabilities and ultimately death. Malfunction of the valves affects their hemodynamic performance in the following two ways:

1. stenosis, i.e., narrowing of the valve and giving a larger resistance to blood flow and a larger pressure drop across the valve.
2. incompetence, i.e., failing to close the valve completely. Blood then flows in the reverse direction (regurgitation) when the valve should be shut.

Both of these conditions reduce the efficiency of the heart and place additional stress and strain upon it.

The decision to perform corrective surgery on the natural valve or replace it with an AHV is often made on the basis of an evaluation of the functional impairment, see table 1.1 for such classifications for aortic valve disease (Wright, 1972). The natural aortic valve will be replaced only if the stenosis or incompetence is classified as severe or moderate.

Table 1.1 Classifications of aortic valve disease

Level of stenosis or incompetence	Mean systolic* pressure drop, mmHg	Incompetence* %
Severe	>100	>60
Moderate	50~100	30~60
Mild	20~50	20~30
Not significant	<20	<20

*Based on a cardiac output of 4~5 L/min.

Despite decades of research and development in the field of AHV, today there still lacks the perfect substitute for the natural heart valve. An ideal AHV should satisfy the main requirements as follows, according to Wright (1972) and Soderlund (1975):

1. Nontoxic and sterile for implantation surgery
2. surgically convenient to insert at the ring structure of the heart
3. showing minimum flow resistance and preventing a significant pressure drop across the valve (less than 20 mmHg)
4. A minimum reverse flow for valve closure and keeping the incompetence of the valve at a low level (less than 20%)
5. low mechanical and structural wear of the valve

6. long lasting (25 years) and its functional performance must not deteriorate with time
7. causing minimum blood trauma so as not to cause anemia
8. showing a low probability for thromboembolism without the use of anticoagulants
9. should not be noisy and disturb the patient
10. should have a modest price

1.2.1 AHV Classifications and MHVs' Characteristics

Currently, approximately 185,000 prosthetic heart valves are implanted each year around the world. Among them, 70% are MHVs and 30% are tissue valves. Although AHV is already considered as a safe and efficient way to replace the malfunctioning cardiac valve, it is not a perfect substitute for the natural valve. Patients with MHV require continuous anticoagulant treatment while tissue valves have the long term durability problem.

Fig.1.4 Carpentier-Edwards porcine valve

Tissue valves are usually constructed by tissue leaflets that are made of biological material, such as porcine, bovine. Fig. 1.4 shows one model of tissue valve known as the Carpentier-Edwards. A major advantage of tissue valve is its excellent hemodynamic performance and hence patients need not take long term anti-coagulation

treatment. Because tissue valves have to undergo glutaraldehyde treatment to kill foreign cells, the implanted tissue valves no longer have the capability of self-repair mechanism. Over time, they either tend to wear, tear or/and calcified, hence leading to failure. This becomes a major disadvantage of tissue valve where it usually lasts no more than 8-10 years.

MHV are made of non-biological biomaterials including pyrolytic carbon (PyC), Stellite, Titanium, silicone rubber, Elgiloy, Teflon, Derin and Dacron. Among them, PyC is the most commonly used material and is well proven. MHV generally has three essential components 1) the occluders or leaflets 2) the cage like superstructure or hinge that guides and restrict occluder motion and 3) the valve base or housing.

Fig.1.5 Photographs of three most commonly used MHVs

The most widely used MHVs are the caged-ball valves, the mono-leaflet tilting disk valves, and the hinged bi-leaflet valves. Typical model of caged-ball valve is Starr-Edwards caged-ball valve as shown in Fig. 1.5 (a). The valve comprises of a polished stellite alloy orifice and a cage that entraps a spherical silicon rubber occluder. The occluder seats directly onto the orifice ring, minimizing leakage through the closed valve. Flow through the open valve passes the silicone ball, which remains in the center of the flow pathway, leaving an annular area available for flow. The presence of the

cage is also likely to increase flow disturbance leading to turbulence during the forward phase.

The second type of MHV is the mono-leaflet tilting disk valves, such as Wada, Lillehei-Kaster, Bjork-Shiley (BS), Medtronic Hall (MH), etc. Fig. 1.5 (b) shows the model of BS mono-leaflet valve. This type of valve has a free rotating disk, whose opening angle varies from 60° to 85° . When the disk opens, it divides the valve area into a major orifice and a minor orifice. In the closed position, the disk sits within the valve orifice supported by the outlet strut. Flow through the valve is distorted by the presence of the occluder and the supporting struts, but the disturbance is believed to be less than the caged-ball valve. However, flow through narrow gaps may lead to high shear stress and such stress can result in blood cell damage and hemolysis.

Bileaflet valves represent the most popular MHV design. In 1977, St.Jude medical (SJM) valve was introduced for clinical implantation, shown as Fig.1.5 (c). Its success resulted in the introduction of other bileaflet valves, such as Edward-Duromedics (ED), Carbomedics, Sorin Biocarbon (SB), ATS, and Medtronic parallel (MP) designs. For the SJM valve, the two semicircular occluders are pivoted to allow free rotation. The wide opening angle and central flow pattern present minimal disturbance to flow. Closing regurgitation is also minimized by the size and shape of the leaflet, however, the central gap between the two leaflets will contribute to regurgitation.

1.2.2 Complications in MHV

An ideal MHV has not developed yet despite decades of efforts to enhance its design and material. The most serious complications are:

- (a) thromboembolism
- (b) hemolysis due to red blood cell destruction

-
- (c) tissue overgrowth
  - (d) infection
  - (e) tearing of sewing suture
  - (f) valve failure due to material fatigue or chemical change
  - (g) damage to the endothelial tissue lining of the vessel wall,
  - (h) leakage caused by failure of the valve to close properly.

Among them, hemolytic and thromboembolic events account for about 70% of all complications associated with MHV replacement, even when strict anticoagulant therapy is followed (Edmunds, 1987). Both of these two complications are directly related to the fluid dynamics and mainly to shear stresses near the surface in the vicinity of the valve.

1) Thromboembolism

The thromboembolism in MHV is quite complex, which is generally believed to be associated with the following factors: (a) blood vessel damage; (b) hemolysis of red blood cell (RBC); (c) the contact of blood with a foreign material; (d) shear induced platelets trauma.

When the endothelial lining of a vessel wall is damaged or eroded off, platelets have access to collagen fibres inside subendothelial connective tissues. The adhesion of platelets to the damaged vessel leads to the release of ADP and the platelet factor 3 (PF-3), which initiates both aggregation and coagulation of platelets.

RBCs contain ADP and a clot-promoting factor known as erythrocytin. When hemolysis occurs, both of these substances are released from a RBC into the plasma, which possibly initiates the platelet adhesion-aggregation and coagulation.

The platelets activation and the coagulation system by contact with artificial surfaces are generally believed to be the key events in the thromboembolic complications for

MHV. Fluid stresses can activate platelets and coagulation at levels on the order of 10 N/m^2 . Hung (1976) reported that platelet damage at $10\sim 17 \text{ N/m}^2$ with exposure time of 100s. Williams (1974) found the same at 13 N/m^2 and 10^{-3}s . Ramstack (1979) reported that platelets are activated at $30\sim 100 \text{ N/m}^2$ and $10^{-2}\sim 10^1\text{s}$. When the platelets undergo relatively high shear stress, they may be damaged with the subsequent release of ADP, which could initiate adhesion and coagulation of the platelets and cause thrombus formation.

2) Hemolysis

Destruction of RBC is generally referred to as hemolysis. Hemolysis can be subdivided into intra- and extra-vascular hemolysis. Intra-vascular hemolysis is the loss of erythrocytes that is sufficiently damaged to cause spillage of hemoglobin into the blood plasma. Extra-vascular hemolysis is the loss of damaged RBCs without lysing in the vascular tree. These damaged RBCs are subsequently removed by the reticuloendothelial system.

Hemolytic disease due to intra-vascular RBC destruction is well documented in patients with AHV and valvular heart disease. A vast majority of the patients with MHV have reduced RBC life span. The bone marrow can, to a large extent, compensate to supply the body with RBCs so that the patient may not become anemia.

Although hemolysis is commonly observed in patients with MHV, it is usually moderate. However, both sublethal and lethal hemolysis may influence the development of a thrombotic state. Lethal red cells damage can cause the release of adenosine diphosphate compound which stimulates platelet aggregation. Continuous sublethal hemolysis can lead to alterations in red cell membrane morphology and reduced membrane flexibility.

Flow properties, most notably turbulence, shear stress, and stagnation, have been

implicated in the blood cell damage and thrombus formation in MHV. Turbulence is generally considered prone to promote thrombus formation or to yield shear stresses capable of contributing to hemolysis (Woo et al., 1985). The extent of blood cell damage is a function of both the magnitude of shear stresses acting on blood corpuscles and their exposure time to the shear field (Hellums et al., 1975).

In assessing the potential turbulent flow to damage blood cells, it is also necessary to study the length scales of turbulent eddies (Kolmogorov scale) as well as the magnitude of the turbulent stresses (Lu et al., 2001). Papoutsakis (1990) and Kawase (1990), after studying the cell damage in bireactors, found that the cell survival rate is closely related with turbulence length scale. If the Kolmogorov scale eddies are of the approximately the same size or smaller than the blood cells, they can be dissipated through interaction with the cell and transfer their energy to the cell membrane, whereas when eddies are larger than the cell, they merely surround the cell to rotate and/or translate it with little relative motion between the cell membrane and the turbulent fluid.

1.3 Particle Image Velocimetry

Experimental method is one important approach to study the fluid dynamics and structure around AHV. With the scientific and technical progress achieved in optics, lasers, electronics, video and computer techniques, more accurate and advanced measuring techniques have been developed, include tomographic interferometry, planar laser induced fluorescence for scalars nuclear magnetic resonance imaging, laser speckle velocimetry (LSV), particle tracking velocimetry (PTV), and particle image velocimetry (PIV). PIV technique was first invented in 1943 by Adrian (Raffel et al., 1998). Early PIV studies relied on opto-mechanical techniques and single frame to obtain velocity data, which made it time consuming and tedious, especially when the

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photographic process was included. The recovered velocity vector from single frame causes ambiguity of the flow direction and lost of phase information. To overcome this problem, image shifting was applied (Adrian 1986a), which further complicated the experimental setup and subsequent data process. In an effort to remove the complex optical analysis process, Blackwelder et al. (1988) digitized photographic records of a flow field and performed the analysis digitally by cross-correlating small sections of two images. Kimura and Takamori (1986) used cross-correlation of video images to obtain the unsteady flow field behind a cylinder, in which two separate images were applied to distinguish the particle images in a double exposed mode.

Currently, digital particle image velocimetry (DPIV) can analyze the digitally recorded video images computationally, allowing the images are captured at real time, viewed and then processed without delay. The directional ambiguity is resolved by implementing cross-correlation technique between two sequential single-exposed particle images. With the help of computer technology and modern video cameras (1000×1000 pixel), more than 100 PIV recordings can be captured per minute. The evaluation of one video PIV recording with 3600 instantaneous velocity vectors (depending on the size of recording and of the interrogation size) is of the order of a few seconds with standard computers.

PIV is an alternative to the conventional hot-wire and Laser Doppler Anemometry (LDA), which measures velocity-time record at a single point in space. The experimental set up of a PIV system normally consists of several subsystems, which work in correct sequence as an integrated one. Tracer particles have to be added to the flow in most applications. Each time the laser is pulsed, the image of the particles in the illumination plane creates an exposure on the camera. The particles inside the flow have to be illuminated twice with a short time interval. The displacement of the particle

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images between the light pulses is determined through evaluation of the PIV recordings. The digital PIV recording is divided into small sub-areas called “interrogation area”. All particles within one interrogation area are assumed moving homogeneously between the two illuminations. The local displacement vector in each interrogation area is determined from the first and second illumination by means of statistical methods (auto and cross correlation). The projection of the vector of the local flow velocity into the plane of the light sheet is calculated, taking into account the time delay between the two illuminations and the imaging magnification. Therefore, PIV is a non-intrusive, whole field measurement technique, which allows its broad application even to high speed flows with shocks or in boundary layers, as well as unsteady flow fields. It should be noted that PIV measures the velocity of a fluid element indirectly by means of the measurement of velocity of a tracer particle within the flow. It is thus required to check carefully whether the particles will faithfully follow the motion of the fluid elements.

Application of PIV technique in cardiovascular engineering is quite new and just began about 15 years ago. For the unsteady, three-dimensional and complicated geometry of physiological flow field, PIV shows its inherent advantage of illustrative fieldwise presentation of flow features continuously in time and space. PIV study of flow around prosthetic heart valves ranges from the investigation of steady flow to pulsatile flow, from enlarged model to real valve, from local visualization such as regurgitant study to full field observations (Lim et al., 1994,1997,1998; Brucker 1997; Zhao et al.,1998; Subramanian et al.,2000; Brucker et al., 2002; Kini et al., 2000; Manning et al., 2003; Balducci et al., 2004; Marassi et al., 2004). These research works greatly helped to understand the flow field and to explain related physiological and pathological phenomenon. However due to the complicated MHV structure and dynamics, also the restrictions of the experimental apparatus, some problems still remain unsolved. For

example, the study for the blood damage in MHV usually focuses on the downstream region (Yoganathan et al., 1986; Lim et al., 1998;) or regurgitant flow (Meyer et al., 1997; Maymir et al., 1998). The flow field and stress distribution inside MHV has not been fully investigated due to the opaque property of the occluders.

1.4 Cavitation in MHV

In vivo cavitation damage of MHV received much attention and interests since the cavitation-like erosion in PyC components was discovered in several ED clinical explants (Quijano 1988; Kleptko et al., 1989). Pitting and cracks on the leaflets and housings were also found in an explanted ED valve from sheep after several month of exercise regimen (Kafesjian et al., 1994).

Cavitation is the rapid formation of vaporous bubbles inside the fluid during the local pressure below the vapor pressure and the subsequent collapses of cavitation bubbles due to pressure recovery (Knapp et al., 1970). Cavitation starts at one of the three suitable nuclei: gas bubbles, solid particles containing trapped gas, or gas trapped in tiny crevices of the container wall. The cavitation bubbles may contain either gas or vapor. A gas-filled bubble requires relatively long period for the gas diffusion into and out. A vapor-filled bubble, on the other hand, needs less time to condense and evaporate. The MHV cavitation is produced by the pressure variations in the flowing fluid due to the occluder closure dynamics. Its duration has been shown to be within order of a millisecond.

An intense shock wave and high speed jets is created upon cavitation bubbles' collapse. It may cause erosion marks to the valve surface. These marks act as potential sites for initial crack growth in the valve body. In addition to damaging valve materials, the pitting or erosion on MHV may become sites of flaw for initial crack growth and

provide rough surfaces for preferential deposition of thrombus. The collapse of vaporous cavitation may also damage formed elements in blood (Paulsen et al., 1999). The rate of plasma-free hemoglobin release was measured under operating conditions to produce vaporous cavitation and a marked increase in the hemolysis rate was observed. A positive relation between cavitation intensity and the rate of hemolysis in vitro is also found (Garrison et al., 1994). Recently, gaseous emboli were detected in patients with implantation of MHV using transcranial Doppler ultrasonography (Brakken et al., 1995; Orsinelli et al., 1994). Gas emboli were implicated in cardiac damage, neurological effects, and degenerative changes of the cervical disks in divers (Lin et al., 2000). It has been hypothesized that the MHV is a source of gas bubble formation and the origin of these gas bubbles may be associated with cavitation (Biancucci et al., 1999).

With the current available technology, it is not possible to directly detect cavitation bubbles in vivo. Although cavitation has been demonstrated in vitro, the cavitation environment does not exist naturally in the human cardiovascular system (Hwang 1998). The mechanism of cavitation on heart valves are not well understood at the present time. Quantitative analyses performed by various investigators, both computational and experimentally, are in agreement that MHV closing actions may cause transient, localized cavitation in the vicinity of the valve. There are two main possible mechanisms for MHV cavitation: water hammer and squeeze flow.

1) Water hammer

Water hammer pressure refers to the pressure change induced by a sudden change of fluid velocity. Suppose a tightly fitting piston is installed on left side of a straight horizontal cylindrical tube, which is full of liquid, as in Fig.1.6. Suppose initially the piston and liquid are moving to the right at a velocity of $u(t)$, then the piston suddenly stops. In the absence of cavities, Freed (1981) reported that the pressure gradient in the

flow direction is calculated as $-\rho \frac{du}{dt}$ and the lowest liquid pressure P_p at the piston surface is given by:

$$P_p = P_a + \rho L \frac{du}{dt} \quad (1-1)$$

Fig.1.6 Schematic of a sudden stoppage piston in tube flow

where $\frac{du}{dt}$ is the changing rate for piston velocity, P_p is pressure on the piston surface, P_a is the ambient pressure, ρ is the density of liquid, and L is the length of the liquid column.

Eq.(1-1) provides a basic understanding of the pressure change generated by the occluder's deceleration, although it can not be directly applied to the case of MHV. In the MHV closing process, the occluder is set in motion by the reverse flow and accelerated to close until it impacts on the housing to stop. The volume of fluid, just above the inflow surface of the occluder is brought to rest, thus experiencing the same deceleration as that of the occluder. A dynamic pressure reduction is created, which may initiate cavitation inception. A sharp negative-pressure spike, immediately following the instant of leaflet closure, can be measured by using a high-frequency pressure transducer.

2) squeeze flow

Squeeze flow, known as squeeze film, refers to the fluid between two moving surfaces.

Cavitation in squeeze film has been the subject of hydrodynamic lubrication in journal

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bearings, rotating rollers, and meshing gear teeth in the industries (Young, 1989).

For the case of MHV, immediately before the occluder impacting on the valve housing, the fluid mass in the gap space between the solid surface of the occluder and housing, is squeezed into motion by the rapidly approaching boundaries (i.e, the leaflet tip and housing stop). This squeeze motion may result in high velocity jet emanating from the gap. The velocity and pressure distributions inside the gap are functions of the local geometry and time. As the jet separates from the channel, it mixes instantaneously with the ambient liquid to form a strong shear layer that induces vortices. The core of the vortex is the low pressure region that can initiate cavitation known as vortex cavitation. The cavitation inception in the vortices centre within the boundary layer has been observed in many hydrodynamics applications (Rood, 1991).

The squeeze flow in MHV was shown to be a localized, rapidly evolving transient phenomenon that lasts for a few microseconds. It has received much attention in recent years due to its importance in MHV cavitation. It is also demonstrated that the geometry of the approaching boundaries, i.e., the leaflet tip and the seating tip of the valve's body, affects the dynamics of squeeze flow (Bluestein et al., 1994; Lee et al., 1994b; Lee et al., 1996; Makhijani et al., 1996). However, detailed investigation of squeeze flow, including direct measurement of velocity distribution and the influence of the valve geometry, still lacks.

1.5 Objective and Scope

Knowledge of the flow field around MHV is of interests due to the link between the fluid dynamics and complications of thromboembolic & RBC damage. Despite intensive studies for the past decades, complete fluid dynamics inside MHV lumen and shear stress distribution in one complete cardiac cycle is still limited.

In-vitro cavitation is generally observed to occur at the last instant of MHV closure. It was hypothesized that MHV cavitation is induced by the abrupt pressure reduction, which is closely related to the squeeze flow phenomenon. The flow physics that may contribute to cavitation is not well investigated although years of research efforts. This Ph.D work aims to study the flow field upon MHV closure, which is believed to help understanding cavitation mechanism. The visualization study in the closing gap of an artificial valve is a challenging experimental task, which is not only interesting for valve studies. Due to the limitation of the PIV system, this study did not extend to the investigation of cavitation dynamics. Hence the scope of this project includes:

- 1) To study the global flow field and related shear forces distribution around a transparent actual size bi-leaflet MHV model inside a dedicated pulsatile mock circulatory loop within one complete cardiac cycle. The simultaneous measurement of occluder dynamics was also carried out.
- 2) To measure the flow inside a gradually decreasing gap during the final journey of MHV closure using PIV technique, thus to investigate the squeeze flow as an indirect indicator for cavitation potential.
- 3) To investigate the effect of valve tip geometry on the squeeze flow and thus the cavitation potential in MHV.

Chapter 2 Literature Review

2.1 Hemolysis in Forward and Leakage Flow of MHV

The threshold values for the lethal damage to RBC vary widely between authors, and depend also on the estimated time it takes for blood constituents to transit through the region occupied by the prostheses. For exposure times typical of passage through heart valves (about 1ms), Sallam and Hwang (1984) showed that turbulent stresses above 400 N/m^2 created by a submerged jet would cause hemolysis. Sublethal damage to red cells can occur at turbulent shear stress levels of 50 N/m^2 (Blackshear, 1972). The shear stress threshold of 150 N/m^2 was considered responsible for lethal damage to red cells under prolonged exposure time (Sutera et al., 1972). Although blood damage by fluid stresses is undoubtedly complicated and insufficiently understood, it appears that stress levels above $150\sim 400 \text{ N/m}^2$ are undesirable.

Many studies have investigated forward flow through an open heart valve in the outflow tract as a source of hemolysis. Yoganathan et al. (1986) conducted the first investigation of peak forward flow fields of MHV under pulsatile condition. By using LDA technique, turbulent shear stress in excess of 120 N/m^2 was detected for SJM bileaflet MHV at 11 mm downstream of the valve. In the case of MH and BS tilting disc valve, the peak downstream stress reaches 200 N/m^2 and 180 N/m^2 respectively.

Using PIV technique, Lim et al. (1998) obtained the velocity vector fields and Reynolds stress mapping in the aortic root region. Four prosthetic valves were tested under quasi-steady flow corresponding to the peak systolic phase. The bioprosthesis showed the highest Reynolds stresses due to the reduction of orifice area, with a magnitude reaching 209 N/m^2 . Lim et al. (2001) continued similar investigation on porcine bioprosthetic heart valve under pulsatile flow condition. The downstream stress mappings were obtained at five different time steps of the systolic phase, which enabled

a time history of the stress level experienced by fluid particles when moving across the aortic root. Thus the propensity of shear-induced damage to platelets and RBCs can be estimated.

The leakage flow after MHV fully closure is another area that attracts much attention. For the small gaps between the leaflet edges and valve housing, high speed leakage jet and elevated turbulent shear are expected to occur due to the large pressure difference across the gap (Ellis et al., 1998). In a two-dimensional, laminar, stationary finite element analysis (FEA) of the SJM MHV, Reif (1991) found that the maximum viscous shear stress through the narrow slit between the closed leaflets is of the order of 700-1000 Pa for a driving pressure difference of 150-300 mmHg. The peak stress also shows dependency on the thickness of the clear space and the velocity in it. Haggag (1990) also employed a theoretical approach to study the leakage shear stress for four AHVs. His results indicated that the average stress was around 270 Pa in the immediate upstream region, regardless of the valve type. Yoganathan et al. (1986) found under aortic pulsatile flow condition, MH tilting disk has the peak reverse velocity of 0.28 m/s and turbulent shear stress of 68 N/m^2 in the area 10 mm upstream of the valve. Baldwin et al. (1994) conducted two-dimensional LDA measurements at 100 near wall positions proximal to the minor orifices of the mitral and aortic tilting disc MHVs in the Penn State Electric Left Ventricular Assist Device. Proximal to the aortic valve, the highest turbulent shear stress was found to reach 990 N/m^2 in the leakage jet and the corresponding peak velocity is 2.8 m/s. Near the mitral valve, the peak turbulent stress is 900 N/m^2 with a peak leakage velocity of 4.4 m/s. These peak turbulent shear stress and velocities were detected 1 mm upstream of the valves and 0.6 mm from the adjacent wall. This study is among the first towards detailed experimental investigation of the pulsating leakage flow in the immediate vicinity of the heart valve. Meyer et al. (1997)

expanded Baldwin's study using three-dimensional LDA technique on similar test setup. The maximum Reynolds shear stress of 1000 N/m^2 and peak velocity of 3.7 m/s were demonstrated for the leakage jet on the mitral position, suggesting insignificant difference from the two-dimensional measurements.

The retrograde flow fields upstream of the hinge area were investigated for two different bileaflet MHVs using 3-D LDA technique (Ellis et al., 1996c). Under pulsatile flow conditions, within the reverse flow phase, the SJM and MP valve demonstrated similar peak velocity and maximum turbulent stress, with the magnitudes around $0.6 \sim 0.7 \text{ m/s}$ and $80\text{-}100 \text{ N/m}^2$ respectively. However, significant difference were detected in the leakage jet. In the case of the SJM valve, the maximum velocity was 0.8 m/s and highest Reynolds stress was 180 N/m^2 , while the MP valve was found with the peak turbulent stress reaching 369 N/m^2 and the peak velocity up to 1.9 m/s . Therefore it was suggested that the retrograde flow field around the hinge is design dependent. Based on this founding, Ellis et al. (1996b) investigated the flow structures at 70 sites inside the hinge region of the valve with the help of a transparent housing. Under pulsatile flow conditions, the LDA measurements indicated that proximal to the housing wall, the peak reverse velocity reached 3 m/s upon closure. This area also experienced the maximum turbulent stress of 600 N/m^2 during leakage flow. The high turbulent stress near the top of the inflow channel combined with a persistent vortex implicated it as a likely region of thrombus formation. This study suggested that hinge design can significantly affect the washout capacity and thromboembolism potential in MP valve.

The effects of the gap width on the leakage velocity and turbulent stresses fields was studied by Maymir et al. (1997). The gap width was deliberately controlled by the temperature variation on the different thermal expansion coefficients of the Delrin occluder and the Stellite housing of BS tilting disc heart valve. Results showed that the

minor orifice tended to form stronger jet than the major orifice during regurgitant flow, suggesting non-uniform gap width around the valve circumference upon fully closure. In the minor orifice of tilting disc MHV, the maximum & time-averaged major Reynolds shear stress and leakage jet velocity increased as the gap width decreased. In the major orifice, however, the opposite trend was demonstrated.

2.2 PIV Technique in MHV Study

Qualitative visualization of physiological flow using imaging and particle seeding has a long history, which can be traced back to the first drawings of Leonardo da Vinci. The reason for particle imaging techniques being well suited in biological flow visualization is the illustrative field-wise presentation of the flow features in the often highly unsteady and three-dimensional flow condition. One special interest is the flow field past AHVs due to the link between flow dynamics and complications of thrombosis & clot formation. For optimization and comparison of MHVs, full field measurements or numerical simulation of the time-dependent flow are imperative. However, the highly unsteady and three-dimensional flow conditions, combined with the complex geometry, elastic walls, moving boundary (valve occluder), makes it an extremely difficult task. The classical point measurement techniques like hot wire anemometry or LDA used for the quantitative in-vitro flow studies did not provide cycle-resolved full field velocity information (Lim et al., 1994). With the computer aided image processing, the PIV technique emerges as a well suited non-invasive measuring technique that provides the instantaneous two-dimensional velocity field throughout a plane wise illumination and offers the calculation of derived quantities such as the vorticity or shear rate (Brucker, 1997).

The PIV studies around AHV normally can be characterized by three categories: 1) enlarged model; 2) specific flow phase of the cardiac cycle; 3) complete analysis of the whole cycle. The application of upscaled model can provide deeper insights of the flow details and enlarged time scales if similarity laws can be satisfied. The earliest velocity field measurements in this respect were carried out by Affeld et al. (1989) in a 10:1 scaled model of a BS monostrut valve and even enlarged model of the ventricle and aorta. Zhao et al. (2001) conducted PIV measurements inside a 3:1 enlarged bi-leaflet MHV model in a pulsatile flow loop. In their study, a cubic flow channel was constructed and the shape of the valve was also rectangular. The results demonstrated strong interaction between the fluid and the moving leaflet. The parameter of d/w was introduced in this experiment, where d was the distance between the two hinge pivots and w was the width of the test section in the middle plane. It was found that the extreme low d/w causes temporal stagnation flow in the central plane, while high d/w tends to produce instability and highly disturbed fluid motion, leading to abnormal opening and closing behaviors. The optimal d/w was suggested in the range of 0.2 to 0.3. As the cubic flow channel and rectangular leaflet applied in this study, which is deviated from the real condition, the results were compromised.

Lim et al. (1997, 1998)) performed the steady flow study around commercial AHVs. The downstream turbulent stress mappings were obtained for four AHVs: a porcine bioprostheses, a caged ball valve, and two single leaflet tilting disc valves with different opening angles under steady flow conditions of 5, 10, 20, 30 liter/min. The porcine bioprostheses exhibited the highest pressure loss and Reynolds stress of all the tested valves. The tilting disk valves, on the other hands, due to its larger opening angle, showed smoother flow profile, lower Reynolds stress and pressure drop. The flow characteristics and shear induced damage for the porcine bioprosthetic aortic valve was

further investigated within systole phase under pulsatile flow conditions (Lim et al., 2001). In this study, the velocity vector fields and Reynolds stress mappings at five different time steps were obtained by PIV technique. The valve exhibited a triangular velocity profile in the downstream for most systolic duration and the acceleration gradually diminished after peak systole. The maximum velocity decreased from 2.3 m/s at peak systole to 1.1 m/s at late systole. High stress regions were produced by the intense momentum exchange between the central jet expansion and the recirculating region inside the sinus. With the knowledge of exposure time and the shear stress level experienced by fluid particles throughout the systole phase, the Lagrangian method was used to estimate the potential shear damage to the platelets and red blood cells.

The flow fields past MHVs in the opening process were investigated by Brucker (1997). In this study, an in-house developed DPIV system was used to study the pulsating flow around MHV. A video-based DPIV with a time-delayed dual-camera recording was built and the delay between the “master and slave” image was arbitrarily adjusted to increase the velocity measurement range. With a frame rate of 50Hz and cross-correlation method, the high dynamic range of the pulsating flow in BS and SB MHVs was captured and the flow evolution process was resolved. The maximum speed of 3 m/s was demonstrated and the vortices, which was generated behind MHV and subsequently swept downstream, were also observed. Brucker et al. (2001) further investigated the flow phenomena and associated occluder dynamics in the systolic phase of a fully transparent 1:1, three leaflets MHV model. The experiment was carried out in the aortic position under pulsatile flow conditions. The velocity profiles at six instants within the systole cycle were measured in both the axial plane and radial cross sections. The occluder opening motion was presented by a high speed camera and subsequent image processing. The phase averaged DPIV measurements indicated that the maximum

axial flow velocity reached 1.5-2 m/s at peak systole, with secondary vortices generating in the vicinity of the occluder. The late systole phase demonstrated decelerating flow near the wall, which finally separated and reversed to move upstream. Besides forward flow phase study, PIV technique was also applied to investigate the regurgitant and/or rebound phenomenon due to its associated cavitation potential and elevated shear stress in MHV. Kini et al. (2000) used PIV to perform a regurgitant study for BS MHV at the mitral position in a flow loop simulating single closure event. This study, in spatial resolution of 1mm, clearly depicted the flow patterns associated with occluder motion, both in the forward closure phase as well as rebound phase. The flow maps in the horizontal central plane and vertical plane of the minor orifice and major orifice side were separately demonstrated for each phase. In particular, flow velocity in excess of 1.5 m/s was detected 3 mm away from the valve seat on the major orifice side created by the occluder rebound, where cavitation has been observed. This study for the first time spatially described the effects of occluder rebound on the flow field around MHV.

Manning et al. (2003) performed a regurgitant study associated with SJM bileaflet MHV on similar test loop. At the instant 5 ms prior to valve impact and 5, 50 and 100 ms after impact, PIV images were acquired at two planes, one is along the valve plane, and the other one is along the central plane, perpendicular to the valve plane. The results indicated the formation of a strong regurgitant jet, in the valve plane, along with two vortices near the leaflet. It was suggested that the near leaflet vortex may provide a low pressure environment for cavitation.

More recently, the flow field investigation within whole cardiac cycle was performed, providing the complete information about MHV operation. Subramanian et al. (2000) carried out PIV measurements of a mitral bileaflet MHV in a pulsatile flow loop. The

transparent MHV model was applied to facilitate the observation inside valve lumen in the opening process, full open phase, closing process and full closed phase. The results indicated that flow in the full open phase formed a three peak axial velocity within three luminal orifices, while the maximum closing flow speed reached 1.44 m/s between the occluder and the housing wall. In particular, the observations of localized jets, steep velocity gradient and vortex recirculation near the MHV surface were suggested to promote platelet aggregation. This study gave a relative complete description of flow evolution inside MHV and was meaningful for the future validation of computational fluid dynamics (CFD) work.

Balducci et al. (2004) used both PIV and PTV techniques to measure the velocity field downstream of a bileaflet MHV within one complete cardiac cycle. Under pulsatile flow conditions, phase averaged flow maps and total stress (shear stress + Reynolds stress) distributions at eight instants were obtained. The PTV measurement gave fluid particle trajectories and the mean time a particle spending in each region, which were combined for the estimation of the stresses on blood cells. However, the mean flow rate applied in this study was only 1 liter/min, which is far below the physiological cardiac output of 5 liter/min.

Marassi et al. (2004) carried out the first three-dimensional PIV investigation upstream and downstream of a bioprosthetic heart valve. The test chamber was a prismatic box with a trapezoidal section to allow optical access for the two cameras as well as to minimize the refraction distortion. The two cameras were carefully stereo-arranged and calibrated. The experiment was performed under two pulsatile flow conditions: one with a mean flow rate of 1.43 liter/min and a pulse rate of 24/min, the other being mean flow rate of 9 liter/min. The complete opening and closing transient of the tested valve was followed and the upstream & downstream 3D velocity maps were presented. Although

results reported in this study were not aimed to provide a complete physiological analysis of the tested valve, the proposed 3D PIV technique is quite unique for the MHV flow field investigation.

PIV sometime is combined with LDA technique to complement with each other and provide an enhanced understanding of the flow around AHV. Browne et al. (2000) investigated the steady flow downstream of the St. Jude bileaflet MHV by using both PIV and LDA technique. Comparisons of mean velocities obtained by the two techniques were in good agreement. For second moment quantities, including the maximum turbulent principle stress and turbulent shear stress, the comparison is less satisfactory. The most obvious possible source of error was due to the insufficient PIV frames of only 32 vs more than 3000 LDA data points at each measurement location. In addition, PIV system used in this study had a spatial resolution of 0.6×0.6 mm, which is typically larger than the LDA sampling volume of $140 \times 20 \mu\text{m}$. They suggested that with rapid increase in computer storage capacity and laser capabilities, these restrictions may be overcome in the near future.

Kini et al. (2001) used both LDA and PIV technique to investigate the transient regurgitant flow downstream of BS MHV in the mitral position. LDA and PIV measurement were taken at both the minor orifice plane (9mm proximal to the valve) and major orifice plane. LDA results showed the fine temporal variation of fluid dynamics associated with occluder first closure and rebound, while PIV gives more intrusive quantitative flow map in the two-dimension plane. The fluid velocity was found to reach 2.0-2.5 m/s at closure and rebound instants for the minor orifice side. And the afterwards sustained minor orifice leakage was measured at 0.5-1.0 m/s for this region. This study suggested that the combination of the PIV and LDA can provide more insight to the complex pulsatile flow past MHV.

Theoretically, CFD results need to be compared and validated with corresponding experiments. PIV, as a fast and convenient measuring technique with adequate temporal and spatial resolution, is often applied for study of unsteady and complex flow in the biomedical engineering area. Bluestein et al. (2000) used both CFD and DPIV technique on the MHV vortex shedding during the deceleration phase. In their DPIV experiment, a prototype bi-leaflet aortic valve was constructed by using a pair of SJM 27 mm pyrolytic carbon leaflets and precisely matched transparent housing ring. The test section mimicked the geometry of aortic root with sinuses of Valsalva machined inside a rectangle flat Perspex. Numerical simulation of the unsteady, turbulent flow through MHV was constructed, using the Wilcox $k - \omega$ turbulence model for internal low Reynolds number flows. The CFD results depicted the intricate pattern of interacting shed vortices in the wake of the valve's leaflet, which were confirmed by the DPIV measurements with good agreements of the spatial and temporal locale of the vortices as well as the velocity profiles in the wake. The results suggested that the shed vortices promote the formation of larger platelet aggregation by the repeated vortex pairing inside the wake and eventually results in free emboli.

2.3 Cavitation and Squeeze Flow in MHV

There are voluminous cavitation studies in the pipeline and turbine machines by using numerical and experimental approaches. The mathematical modeling by characteristics approach can predict closely the pressure transients for vaporous cavitation in the transmission lines (Shu, 2003). But the comprehensive analytic predictive method is prohibited by the complexity of the solution to the complete Navier-Stokes equations for two phase flow. Hence the investigations of real fluid effects have been limited to experiments. Previous studies show that the presence of the flow separation, geometry

of the separation region, as well as the surface roughness and flow structure interaction, etc, contributes to both the form of cavitation and the inception conditions. The quantitative measurement techniques are however challenged by the three-dimensionality and short time duration of this event.

Cavitation in MHV was first highlighted in the late of 1980's with the structural failure of the Edwards-Duromedics valve. The MHV cavitation studies in recent years can be divided into two basic categories. One category focused on the observation of cavitation bubbles in various flow systems with different testing conditions, together with the determination of the loading threshold (Graf et al., 1991,1992,1994; Lamson et al., 1991; Zapanta et al., 1994; Shu et al., 1994; Lee et al., 1994b,1996; He et al., 2001). The other category, however, emphasized the flow dynamics of MHV closure and associated flow phenomenon. The factors that may influence MHV cavitation were generally or individually addressed in these studies (Guo et al., 1990; Gross et al., 1991; Wu et al., 1994a,1994b,1995a; Bluestein et al., 1994; Makhijani et al., 1994,1996a,1996b; Lee et al., 1994a; Garrison et al., 1994; Sneckenberger et al., 1996; Chandran et al., 1997,1998; Lim et al., 2003).

For the studies in the first category, in-vitro MHV cavitation investigation often applied optical visualization methods for the observation of bubble dynamics in flow systems simulating the pulsatile flow of the left heart and/or the single closure of the mitral valve. Graf et al. (1991) studied cavitation potential of ten different types of MHVs at the mitral position in a pulsatile circulatory test loop. In this study, the systolic duration was maintained at 15% of the cardiac cycle time and the heart rate was kept at 70 beats/min. An experimentally measured maximum ventricular pressure slope (dP/dt) was used as the loading index. The valve's inflow surface was illuminated by triggered strobe light and imaged by a video camera. Cavitation threshold was determined by the

left ventricular dP/dt value at which the first cavitation bubbles can be detected by the video camera. All the valves demonstrated cavitation upon dP/dt value rising up to 5500 mmHg/s. Some types of valves were found to have lower cavitation threshold values than the physiological dP/dt range of 500~2000 mmHg/s, which is the normal range for a healthy human at resting condition.

Similar works was carried out by Graf et al. (1992) under simulated normal physiological resting and exercise conditions. In this study, the cavitation threshold demonstrated strong dependency on the heart rate (i.e. resting or exercise condition). None of the tested MHVs showed cavitation under normal physiological resting conditions. However under exercise conditions, even lower dP/dt values of 2000~3250 mmHg/s can induce cavitation occurrence for most of the valves.

In the above-mentioned studies, the maximum left ventricular dP/dt was used as a loading index for the determination of MHV cavitation potential. This method has been questioned due to the fact that the maximum left ventricular dP/dt can only occur after the complete closure of the valve. In addition, many other factors, such as mounting compliance, valve size and specific design feature, etc, other than dP/dt value, were also found to affect MHV cavitation. Lee et al. (1994b) proposed to use the transvalvular (ventricular minus atrial pressure) pressure rise rate averaged during valve closing period as the loading index for MHV cavitation evaluation. The start and ending instants of valve closure were determined from two monitoring pressure traces in a flow system simulating single event of mitral valve closure. In vitro cavitation studies were performed on nine different MHVs by observing the bubbles on the valve inflow surface. The inflow surface was illuminated by a triggered stroboscope and photographed by a camera. Cavitation bubbles were observed around the seat stop, along the inflow strut, along the peripheral clearance or edge of the occluder, and along

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the central clearance, with demonstration of clearly defined bubble fields of MHV cavitation at various time delays after the occluder/housing impact.

Wu et al. (1995c) investigated the relationship between the ventricular pressure slope and leaflet impact velocity in the mitral position of a mock pulsatile flow loop. This study showed that the left ventricular slope (dP/dt) under various heart rates and cardiac outputs are dependent to the valve type and given driving condition of the left ventricle. It was suggested that the maximum ventricular dP/dt , which occurs after the valve complete closure, should not be used as a reference to study mitral valve closure.

Chandran et al. (1998) obtained in vivo dP/dt value ranged from 700 to 2300 mmHg/s for two MHVs and one tissue valve when they were implanted in the mitral position of a goat and infused with certain medicines to simulate hypo- and hyperdynamic conditions. The in vivo dP/dt , which was averaged from the ventricular pressure rise rate during valve closure, agreed with the loading magnitude computing with the same method in vitro experiments. They therefore suggested that it is appropriate to employ averaged ventricle pressure slope as the loading index for MHV cavitation study.

Based on these earlier MHV cavitation studies, the US Food and Drug Administration (FDA) drafted the guidelines for the evaluation of MHV cavitation potential (Herman et al., 1994). The current FDA guidelines define MHV cavitation threshold level as the ventricular pressure slope (dP/dt_{20}) 20 ms prior to the valve closure by direct observation of bubbles on the MHV surface. An interlaboratory comparison of the FDA protocol was carried out for cavitation threshold measurements on a common set of six clinical MHVs (Carey et al., 1995). Five different industrial and government laboratories were involved, from which significant differences existed in the measured MHV cavitation threshold (dP/dt_{20}). This may result from the differences in test

systems, including a wide range of ventricular compliances and single valved vs double valved test systems.

Factors affecting MHV cavitation and associated flow characteristics have also been extensively studied in the past decade. MHV occluder closure behavior has widely believed to be the key factor in MHV cavitation. Transient pressure reduction due to the occluder's sudden deceleration at valve closure, regurgitant and squeeze flow ejecting from the gap have also been suggested contributing to the cavitation inception in MHVs.

Guo et al. (1990, 1991) developed a laser sweeping technique (LST) to measure leaflet closing motion within its last three degree of closure journey. By using this technique, the MHV occluder closing motion, with a few milliseconds prior to the leaflet/housing impact, can be measured with a few microseconds sampling time interval. Several types of MHVs' final closing behaviors were characterized by using this technique (Guo et al., 1991; Wu et al., 1994a,1994b,1995a,1995b). Occluder impact velocities of various types of MHVs were found ranging from 0.9m/s to 5.0 m/s at heart rates from 70 beat/min to 140 beat/min. The occluder closing motion at final stage exhibited a rapid change and generally consisted of three phases: 1) occluder approaching; 2) occluder deceleration; and 3) occluder rebounding.

Gross et al. (1991) performed a LDA measurement of the leakage jet velocity to study Venturi effect on MHV cavitation inception. In a steady testing flow chamber with different transvalvular pressure drops, the leakage velocity through various central leakage gaps between two leaflets showed a geometric independence. It indicated that the Venturi pressure can not cause MHV cavitation in a closed valve, which basically rules out the occurrence of cavitation after valve fully closure.

Bluestein et al. (1994) carried out CFD study for the closing velocity distribution in the gap flow field of a bi-leaflet MHV in the mitral position. With the valve leaflet's geometry and constant closure velocity, 4m/s as the input, computations were carried out to determine velocity distribution in the gap flow field using commercial software FIDAP. Their two dimensional, time dependent simulation showed that the flow velocity in the gap space can reach values as high as 30m/s in regions near the inflow surface of the ED leaflet. According to the Bernouli principle, this is a magnitude nearly three times higher than that required to vaporize water under standard atmospheric pressure, and can create the conditions that characterize cavitation. Additional computational work also showed that the squeeze flow was mainly influenced by the leaflet motion and geometry of the gap channel rather than the atria-ventricular pressure difference and fluid viscosity.

Makhijani et al. (1994) also performed a similar CFD numerical simulation of the squeeze flow field in the gap of ED MHV. The squeeze flow velocity reached a relatively low value (about 9m/s) after considering leaflet deceleration and rebound. However, the minimum fluid pressure in the computation dropped below the vapor pressure of the liquid during the leaflet deceleration and rebound phase. They later coupled CFD simulation with occluder motion and housing deformation in response to the surrounding fluid (Makhijani et al., 1996a). And it was shown that the valve mounting compliance, in addition to the geometry of the contact region, also influenced the squeeze flow field. Siegel et al. (1996) simulated two-phase flow through the 2D MH valve. They found for a rigid mounting, a vapor region was formed immediately distal to the squeeze flow gap region. But for a flexible mounting of the same valve, results showed a significant reduction in the vapor volume through valve closure and rebound. They also compared the flow fields surrounding a bi-leaflet MHV with the

housing and without the housing using CFD technique (Siegel et al., 1997). The results indicated that peak squeeze flow velocities were significantly reduced from 4.3 m/s with the valve housing to 2.1 m/s without the valve housing. The peak vapor volume was also reduced five-fold times with the removal of the housing. This study provided an assessment of the relative importance of squeeze flow to other fluid dynamics effect, such as water hammer for the mitral MHV closure phenomenon.

Chandran et al. (1994) experimentally gave a rough estimation of the squeeze flow velocity of 12 m/s in a flow loop simulating single closing of mitral valve. The cavitation bubble was imaged in the inflow side and the squeeze flow velocity was calculated based on the time of the pulses and the distance between similar bubbles, assuming the cavitation bubble was caused by squeeze flow and the same bubble's radial motion was followed. Maymir et al. (1998) found fluid velocity spike in the minor orifice of BS monostrut valve upon closure, with a magnitude of 1.67 m/s in the aortic position and 3.02 m/s in the mitral position when they carried out two-dimensional LDA measurements of velocities and turbulent stresses inside a ventricular assist device. It was suggested that these velocity spikes were the evidence of "squeeze flow" since they were significantly larger than the corresponding occluder tip velocities upon closure and only lasted only one sample interval----~1ms.

Lee et al. (1994b) identified possible valve design factors for the cavitation initiation when studying nine different MHVs. The occluder stop seat, where fluid squeezing effect may occur, was suggested to be most critical to cavitation since the valve with this particular feature showed a high intensity of cavitation and threshold loading rate below the normal reference physiological value (750 mmHg/s). They later performed a further investigation of transient pressure measurements and cavitation bubble visualization at seven locations of a MH MHV in a flow loop simulating single closure

event (Lee et al., 1996a). The presence of pressure falling below liquid vapor pressure and appearance of bubbles around the seat stop suggested that the fluid squeezing between the stop and occluder was the main cause of cavitation in MH valves. The threshold loading rate for cavitation initiation around the stop was found to be very low (300~400 mmHg/s). This suggested that MHV with stop was more vulnerable to cavitation.

He et al. (2001) indicated that in terms of shape and location of the imaged cavitation bubbles occurring at various time instants, the MH MHV cavitation in a pulsatile mock test loop was shown to be induced either by squeeze flow jet, the transvalvular jet, the tension wave or a combination of the above.

More recently, Lim et al. (2003) used the closing dynamics of two impinging rods to study the MHV cavitation with respect to squeeze flow. The velocity of the movable rod was adjusted by the input pressure and calculated from a series of images captured by high speed camera. The fluid squeezing between the stationary rod and the movable rod was estimated by lubrication theory. Accompanying with cavitation bubble visualization, the effect of rod impact velocity, contact area and squeeze flow on the cavitation inception was studied. This study indicated that the decrease of the contact area and squeeze flow velocity delayed the onset and intensity of cavitation.

Transient pressure at MHV closure, which is a direct result of fluid mass deceleration at valve closure, was found to correlate with cavitation bubble inception and bubble collapse. Chandran et al. (1994) and Garrison et al. (1994) measured the transient pressure at MHV closing by using high frequency pressure transducers while simultaneously observing cavitation bubbles in in-vitro flow loops. Both studies demonstrated large negative pressures transient (NPT) accompanying local bubble formation on the inflow side at the instant of valve closure. They suggested that

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transient pressure drops were correlated with the cavitation bubble inception and collapse.

Chandran et al. (1998) performed NPT correlation study between in vivo and in vitro for the implanted bileaflet MHV in the mitral position. They found that the NPT signal and corresponding power spectral density plots from the in vivo data were in qualitative agreement with those recorded in vitro under the loading rate of 1500 mmHg/s. This study for the first time demonstrated the presence of NPT in vivo, whose magnitude was below vapor pressure in blood, suggesting a cavitation potential in vivo with rigid occluders of MHV. However, the difficulty of precise fixture of transducer inside animal model at the vicinity of the valve may pose variations of NPT recordings in vivo.

Garrison et al. (1994) observed high frequency pressure oscillations (HFO) about 1ms after the pressure drop by placing a high frequency transducer at a location 4.5 cm proximal to the valve surface. They suggested that the HFO were impulsive pressure arising from the cavitation bubble collapse, based on the optical visualization of bubble collapse. Wu et al. (1995a) made a detailed experimental study of the transient pressure at the closing of a Medtronic Hall MHV in a deliberately controlled physiological flow condition. The transient pressure and the closing velocity were simultaneously measured and strong correlation between these two sides was demonstrated. They also observed bursts of HFO after transient pressure reduction. The pressure reduction and the HFO were locally generated proximal to the valve and can be characterized in both time and spatial domains. The high-frequency pressure fluctuation in vivo was studied by Paulsen et al. (1999). Their study indicated high-frequency pressure fluctuation in patients with implanted MHV by using a hydrophone, suggesting a possible cavitation bubble collapse in vivo.

Valve mounting compliance was considered as a possible important factor in MHV cavitation. Kafesjian et al. (1990) indicated that the compliance of the sewing ring may significantly affect the closing force and cavitation bubble formation in ED MHVs. They found that cavitation bubbles can be totally eliminated by increasing the compliance of the sewing ring. Wu et al. (1994a, 1995a) showed that the mounting compliance played an important role in MHV occluder deceleration and rebound, NPT pattern and subsequent HFO with a MH MHV in a pulsatile mock flow loop. Guo et al. (1994) performed a detailed investigation of the effect of structural compliance on MHV cavitation threshold measurements. A specially designed compliance adjustable fixture was used to hold the test valve so that the valve mounting compliance could be measured during the cavitation threshold measurements. The results indicated that the mounting compliance did affect the cavitation threshold. However, in the experiment performed by Lee et al. (1996b), no qualitative difference was found for the pressure transient as well as cavitation bubble visualization under different valve holders in the mitral position inside a single closing event flow loop. Chandran et al. (1997) also did not observe any significant difference of measured initial valve closure velocity with the three different holders in the same test loop. The correlation between NPT measured in vitro and in vivo provided evidence that the flexibility of valve housing was not an important factor for MHV cavitation in vivo (Chandran et al., 1998). Therefore some investigators suggested that the MHV occluder impact velocity upon closure and initial NPT was dependant on the relative motion of the leaflet to the valve housing, which was anticipated to be independent of the compliance characteristics of the housing and its attachments. The importance of the valve holder flexibility for MHV cavitation needs to be investigated further.

Generally speaking, PIV investigations in MHVs have been extensively performed in steady and pulsatile flow conditions, which are instrumental in coupling the fluid dynamics of the prosthetic valve designs to their observed clinical problems, i.e., increased propensity for hemolysis and thrombus formation. The flow behavior upstream or downstream of MHVs, as well as shear-induced damage in the forward and leakage flow, was well documented in many works. However, the study about the flow within a valve lumen needs to be further carried out, in particular, the simultaneous measurement of the valve leaflet dynamics and the nearby flow patterns. The complete knowledge about the variations of the stress distribution within one complete cardiac cycle is still lacking.

MHV cavitation is generally observed to occur at the last instant of MHV closure. For the past few years, several groups have performed extensive studies of in-vitro MHV cavitation. The appropriate dp/dt for cavitation threshold was determined, along with the bubble visualization using stroboscope and high-speed camera. The mapping of the transient pressure at valve closure was found to correlate with cavitation bubble inception and collapse. LST technique was specially developed to measure the occluder closing velocity within the last 3°. The detailed leaflet closing velocity was used for the CFD model for the squeeze flow computation. However significant discrepancies existed among CFD study due to the complicated transient and fluid-structure interacted flow features within the tiny channel. Thus experimental investigation remains to be fully carried out, in particular for the relationship between valve design, squeeze flow and cavitation potential.

displacement) is adjusted to produce the optimal physiological pressure and flow rate in the test loop.

Fig.3.2 Picture of experiment setup

Fig.3.3 Simulated piston motion curve

The piston head is connected to a Perspex box. The box is full of liquid, inside which a collapsible ventricular sac (VC) is installed. The VC is made of Silastic and moulded to

mimic the geometry of a human left ventricular chamber during diastole. Upstream of the VC, is the left atrium chamber connected via a commercial tilting-disk valve. Downstream of the VC, is an in-house developed aortic valve model installed inside a test section. The atrium chamber is opened to the atmosphere.

The cyclic back and forth movement of the piston “dilates” and contracts the VC, which simulates the periodical motion of the left ventricle. In the systolic phase, the piston pushes the fluid inside VC to pass through the aortic valve and force the valve open, while the mitral valve is closed within this duration. In the diastolic phase, the piston moves back to “dilate” the VC, the backflow across the aortic valve forces it to be closed, while the mitral valve is pushed to open, and the fluid from the atrial chamber fills the VC.

The transparent testing section, illustrated in Fig.3.4 and Fig.3.5, simulates the aorta part. The transparent tube is covered with rectangular box to minimize the light scattering in PIV experiment.

Fig. 3.4 Schematic drawing of the test section

Fig.3.5 Test section installed with transparent MHV model

Inside the test section, an in-house developed transparent 1:1 bi-leaflet MHV model is installed, where the flow field measurement is carried out. The aortic MHV model has the similar geometry and opening & closing angle as that of SJM 29 mm MHV, as shown in Fig.3.6. The MHV model was fabricated in order to facilitate the flow field observation inside the two occluders as well as the gap flow between the occluder and

the housing wall. The transparent valve model was made of Perspex, which has a density of 1.2 g/cm^3 , different from commercial MHV of $1.5\sim 2.0 \text{ g/cm}^3$.

Fig.3.6 Transparent 1:1 bi-leaflet MHV model

The afterload system composes of two compliance units and two resistance elements. All of them can be individually adjusted to simulate the systemic impedance and maintain the pressures and flow rate in the physiological range. In the modeling studies of the physiological impedance of the systemic circulation, the nonlinear resistance and compressibility of the circulation system are normally linearized to the resistance R and capacitor C , thus electrical-fluid analog method could be applied. In this test rig, by trial and error, RCRC model was finally adopted based on Windkessel model of RCR.

3.2 Physiologic Pressure Measurement

Several factors, including the piston motion curve, the compliance and the resistive values of the mock circulation loop, affect the pressure and flow rate waveforms in the flow system.

Two pressure transducers (T4812AD-R, Viggo-Spectramed, Oxnard, CA) were used to measure the ventricular and aortic pressures in the test loop. The ventricular pressure

was monitored at the immediate exit of the left ventricle, while the aortic pressure tapping point located six times downstream of aorta diameter for the complete recovery of aortic pressure. The two channel pressure signals were first amplified and filtered by an amplifier (AM-PACK AP9991, VIVITRO SYSTEMS, CANADA) and then simultaneously recorded by an oscilloscope (Tektronix, TDS36). The amplifier was calibrated prior to use and its calibration was accomplished by adjusting its gain & shift with the pressure transducers connecting to a U shape mercury manometer and an oscilloscope. One typical calibration curve is shown in Fig.3.7, with the resolution of 25mmHg/V. The amplifier needs to be re-calibrated every three months to avoid signal shifting.

Fig.3.7 Amplifier calibration curve

Fig.3.8 shows the ventricular–aortic pressure waveforms for the transparent MHV model inside a working fluid of blood analog. The blood analog, whose description will be discussed in the later section, was made up of glycerin, distilled water and sodium iodide to simulate the viscosity of human blood as well as to eliminate optical deformation in PIV experiments. The ventricular and aortic pressure recorded across the MHV model, as in Fig.3.8, fall in the range of 0~165 mm Hg and 60~130 mm Hg

respectively. The systolic duration is 40% of the heart period and heartbeat is maintained at 60 beats/min. The flow rate was recorded by a calibrated rotameter (Blue White Inc, USA) at the location where the fluid returns back to the atrium chamber. The flow rate was adjusted between 3.2~4.3 liter/min. The selected heart beat and flow rate were a bit lower than normal physiological values in order to protect the transparent MHV model for long operation.

Fig.3.9 is the reference physiological pressure waveform from British Standard.

Fig.3.8 Typical ventricular and aortic pressure waveforms

Fig.3.9 Diagrams of Haemodynamic Waveforms Simulating Healthy Humans (from British Standard BS EN 12006:1999)

3.3 Particle Image Velocimetry (PIV)

3.3.1 PIV Setup

The PIV system used in this investigation is the TSI PowerView™ PIV system (TSI Incorporated, USA). It consists of illumination system, image capture system, synchronizer and analysis & display system, as shown in Fig.3.10.

Fig.3.10 PIV hardware setup

1) Illumination System

The illumination system, which includes laser system and lightsheet optics, produces double-pulsed light for the flow region illumination. With the integrated beam optics, the two laser pulses from the two separate Nd:YAG lasers systems (MiniLase III, New Wave Research, U.S.A) are overlapped precisely to form single beam of a co-axial pulse. The use of dual lasers enables the time between laser pulses as short as $0.3\mu\text{s}$, without sacrificing pulse energy. A flashlamp produces the energy for laser beam, whose energy can be adjusted by setting the Q-Switch delay time. The laser pulses have

high energy (50mJ per pulse), and short duration (5-7ns) during the pulse exposure. The laser operates with a frequency (repetition rate) of the double pulses at 15 Hz, which allows continuous capture of image pairs rate of 15 Hz when the PIV camera operating at 30 frames per second.

The lightsheet optics is used to shape the laser beam into a planar light (lightsheet) to illuminate the fluid field. A negative (concave) focal length lens (cylindrical lens) diverges the beam in the height direction, while a spherical lens is used to control the thickness of the lightsheet. The shorter and/or the thinner of the lightsheet, the higher light intensity is acquired. The camera is usually focused near the waist, shown in Fig.3.11, where the lightsheet has a minimum thickness and the strongest intensity.

Fig.3.11 Lightsheet optics dimensions
(from PIV Instruction Manual: Theory of Operation)

Table 3.1 Lightsheet height (mm) and thickness (μm)

Spherical lens FL(mm)		200	500	1000
Cylindrical lens FL(mm)	-15	37(H), 42(T)	97(H), 106(T)	197(H), 212(T)
	-25	21(H), 42(T)	57(H), 106(T)	117(H), 212(T)
	-50	9(H), 42(T)	27(H), 106(T)	57(H), 212(T)

‘H’ denotes height, ‘T’ denotes thickness

For the current PIV system, table 3.1 lists the height and thickness dimensions of lightsheet under various combinations of focal lengths (FL). The selection of these two optics lens was based on the configurations of the test rig and dimensions of the illumination plane, etc.

2) Image Capture System

The image capture system includes a solid state charge couple device (CCD) camera (630046 PIVCAM 10-30) mounted with a Nikon Micro-Nikkor lens f/60mm or a microscope (Stemi 2000-C, Carl Zeiss) depending on different magnifications required. The camera offers $1K \times 1K$ pixel resolution combined with 30 frames per second, which allows it to provide PIV measurement with high accuracy and high image capture rate. The frame straddling technique delays the first laser pulse till the end of the first frame, while the second laser pulse is captured at the second frame. The value of pulse separation time (dT), which can be down to less than $0.3\mu s$, can be adjusted for flow velocities from mm/s to supersonic speed. Another advantage for the camera is its progressive-scan, interline transfer technology, which reduces the "read-out" time between successive frames down into the micro-second order and thus increases the velocity measurement range.

When the camera operates in the frame straddling mode, longer exposure time is needed for the second frame since the first $255\mu s$ exposure should be transmitted before the second image can be moved into the readout register. Therefore it is important to minimize the surrounding light so that no other extraneous light would be captured on the second frame during this relative longer exposure time.

High speed frame grabber is needed to transfer the image from the camera to the host computer at the full camera frame rate. Two images are then analyzed using

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cross-correlation algorithm for the two-dimensional flow velocity computation.

3) Synchronizer

The synchronizer (model 610032) communicates with the host computer by RS-232 commands. It provides the correct timing and the sequence for the laser and camera operation, so that PIV illumination and image capture components are integrated together and function as an automated system.

The synchronizer receives commands from an application program running on the host computer for parameter setting of laser pulse energy, pulse separation and delay time. The exposure and capture mode for the CCD camera is also controlled by this program. Trigger signal outputs from the synchronizer control the Nd:YAG laser pulsing sequence so that the laser pulses are located in the appropriate frame(s) with a very short time interval between-frames, using a method known as frame straddling.

For the detailed functions and connections of the input and output ports on the back panel of the synchronizer, see Components Installing Manuals.

4) Computer and Software

INSIGHT™ 2 version 3.0 software package was installed in a Pentium III computer (450 MHz CPU, 256 MB RAM) under Windows NT version 4.0. The software automates the capture, display and processing of large number of PIV images. INSIGHT 2 performs the image evaluation by dividing each image into many small interrogation regions, within which cross-correlation algorithm is carried out to determine the particle displacement. The particle velocity in every interrogation region is then computed, based on particle displacement and time between laser pulses.

A. Cross-correlation Algorithm

Particle images techniques usually have 3 types of image density. In the case of low image density, matching pairs of particle images from different illumination is feasible and the displacement is directly measured between images of each pair. This method is referred to particle tracking velocimetry, or PTV. For the high image density, the individual particle image overlaps to produce speckle pattern, for which Young's fringe method is used to extract the speckle motion. This situation is often called laser speckle velocimetry, or LSV. For most experimental practices, the concentration of scattering particle in PIV experiment is between the above-mentioned scenarios, i.e., the image of individual particle can be detected but tracking individual particle is impossible or/and time consuming. Statistical approaches are therefore applied to probabilistically infer the proper pairings and measure the average displacement of the group of particles in the interrogation cell. With the improvement of computer speed and storage capacity, the interrogation of the PIV recording is more commonly performed fully digitally instead of using optical Fourier transform. Fig.3.12 illustrates the relationship between a pair of successive particle image .

Fig.3.12 Digital model for the relationship between successively recorded particle image frames (From Raffel et al. Particle Image Velocimetry: a Practical guide, 1998)

The function of interest is the displacement vector $\mathbf{d}(i,j)$, which is nothing more than a Dirac delta function $\delta(i-x, j-y)$ displaced x,y units from image 1 to image 2. The

location of this displacement delta function is the maximum of the two-dimensional spatial cross-correlation $R_{II}(x, y)$ if neglecting the noise effect

$$R_{II}(x, y) = \sum_{i=-k}^k \sum_{j=-L}^L I(i, j) I'(i+x, j+y) \quad (3-1)$$

Where $I(i, j)$ and $I'(i, j)$ are the intensity values of the two digitized interrogation regions.

The peak correlation value corresponds to the correlation of images of particles from the image 1 with images of identical particles from the image 2, and representing the average displacement from image 1 to image 2. The other correlation values are the correlation of different random distributed particles and therefore mainly the noise in the correlation plane. Caution should be taken that cross-correlation method inherently recovers average linear shifts only between two particle image samples, without any rotation or deformation. Therefore, interrogation window size should be chosen sufficiently small to avoid the displacement gradient effect.

The digital PIV evaluation allows a subpixel measurement of the correlation peak using various estimator methods. Generally Gaussian peak fit curve as

$$f(x) = C \exp\left[\frac{-(x_0 - x)^2}{k}\right] \quad x_0 = i + \frac{\ln R_{(i-1,j)} - \ln R_{(i+1,j)}}{2 \ln R_{(i-1,j)} - 4 \ln R_{(i,j)} + 2 \ln R_{(i+1,j)}} \\ y_0 = j + \frac{\ln R_{(i,j-1)} - \ln R_{(i,j+1)}}{2 \ln R_{(i,j-1)} - 4 \ln R_{(i,j)} + 2 \ln R_{(i,j+1)}} \quad (3-2)$$

is implemented, since the well focused particle image itself can be approximated by a Gaussian intensity distribution and the correlation peak is also roughly the Gaussian shape.

B. Validation Filters

After the initial processing, over 95% PIV velocity vectors are correct ones. Next, four kinds of filters, namely global range filter, local median filter, local mean filter and smooth filter, are applied to validate the velocity vectors and remove the incorrect ones.

Global range filter removes the obviously impossible vectors in terms of the maximum and minimum user input values of U and V components of the velocity, while median and mean filter eliminate the vectors that have abnormal change gradient from the neighboring vectors. Smooth filter should be used with caution since it replaces each velocity vector with a weighted average of itself and neighboring vectors. For a carefully designed PIV experiment, normally only the global range filter is used to validate the vectors, while mean and median filters, especially the smooth filter is seldom applied.

After evaluation and filtering, the obtained vector files are loaded into Tecplot software for graphically presentation of the PIV field data, including vector map, vorticity and strain rate, etc.

3.3.2 Parameters Considering for PIV Experiment

In order to optimize PIV performance, a number of parameters should be carefully selected in the experimental process, including magnification, pulse separation time, particles imaging, image concentration, etc.

1) Coefficient of Magnification

Magnification (M) is defined as

$$M = \frac{ObjectSize}{ImageFieldSize} = \frac{FOV}{CCDSensorSize} \quad (3-3)$$

Where field of view (FOV) is represented by the horizontal or vertical length the camera

digitizes in the object plane.

$$\text{Spatial resolution} = \frac{l_{\text{image}}}{M} = \frac{9\mu\text{m} / \text{pix} \times 64\text{pix}}{M} = \frac{576\mu\text{m}}{M} \quad (3-4)$$

where l_{image} refers to interrogation size.

The interrogation area is normally overlapped by 50% under Nyquist Sampling criterion, the velocity vector spacing is therefore half of the spatial resolution. Table 3.2 shows parameters including FOV and velocity spacing under different magnifications.

Table 3.2 FOV and velocity spacing under different magnifications:

		M	FOV (mm)	Velocity spacing (mm)
Micro-Nikkor lens		0.28	31.8×32.3	1.03 mm
Microscope	Zoom=1	1.8	4.83×4.91	0.16
	Zoom=1.6	2.6	3.41×3.46	0.11
	Zoom=2	3.2	2.69×2.73	0.09

2) Pulse Separation Time

Normally, pulse duration time belongs to laser's intrinsic characteristics and already set up in the factory. It is often short enough (less than 10ns) to freeze the particle motion at one instant and therefore avoid image streaking. But the time interval between laser pulses should be chosen according to different application requirements, which is normally satisfied by the following formula, i.e. the displacement between two consecutive frames should be less than one forth of the interrogation size l_{image}

$$V_p \times t_{\text{sep}} < \frac{1}{4} \times \frac{l_{\text{image}}}{M} \quad (3-5)$$

where V_p is particle velocity and t_{sep} is pulse separation time.

In the following experiments, Micro-Nikkor lens and microscope with different zooms

were used to observe continuously smaller flow field. Therefore the pulse separation time was necessarily adjusted for each setting due to the change of both the maximum flow velocity and the magnification.

3) Particle Image

The actual image recorded on the CCD camera is the convolution of the diffraction-limited image, d_s of the recording optics, with the geometric image Md_p , and the effective particle diameter is given by:

$$d_e = [M^2 d_p^2 + d_s^2]^{1/2} \quad (3-6)$$

$$d_s = 2.44(M+1) \frac{\lambda}{2NA} \text{ or } d_s = 2.44 f^\# (M+1) \lambda \quad (3-7)$$

where

d_e : Effective image diameter of particle

d_p : Particle diameter, $10 \mu\text{m}$ for silver coated glass sphere in this experiment

d_s : Diameter of diffraction spot

NA : Numerical aperture of the microscope objective,

λ : Wavelength of the recording light, 532 nm ,

$f^\#$: f number, the ratio between the lens' focal length and the aperture diameter

NA determines the lens's ability to gather light and resolve fine specimen detail at a fixed object distance, which is defined as:

$$NA = n \sin \beta \quad (3-8)$$

Where n is the refractive index of the immersion medium adjacent to the objective and β is the angular radius of the collection cone. For the special case where the medium surrounding the lens is air ($n=1$), and the object distance is large compared to the

aperture of the lens, NA can be approximated by

$$NA = (2 f^\#)^{-1} \quad (3-9)$$

The minimum particle image should be larger than the pixel dimension ($9 \mu\text{m} \times 9 \mu\text{m}$), and if the particle image diameter is resolved by 3-5 pixels, the location of a particle image correlation peak can be determined to within $1/10^{\text{th}}$ the particle-image diameter (Prasad et al., 1992). Thus the measurement uncertainty Δx can be described as

$$\Delta x \approx d_e / 10M \quad (3-10)$$

Table 3.3 Particle image diameter and measurement uncertainty

$\lambda = 532 \text{ nm}, d_p = 10 \mu\text{m}$		image diameter d_e (μm)	measurement uncertainty Δx (μm)
Micro-Nikkor lens $M=0.28, f^\# = 16$		27	9.6
Microscope, NA=0.07	M=1.8	32	1.8
	M=2.6	42	1.6
	M=3.2	50	1.56

Table 3.3 lists particle image size and measurement uncertainty calculated from equation (3-6) and (3-10). It can be seen that under different magnifications, the particle image sizes basically fall into the optimized range and thus justify the particle selection and magnification setting.

4) Particle Requirement

Two requirements need to be considered when selecting the seeding particle. The first one is that the particle should be small enough and buoyancy-median to faithfully follow the original flow, while on the other hand, the particle should be big enough to scatter adequate light for image detection. The optimum particle size is therefore a

compromise between these two requirements so that the high resolution particle image can be obtained. The basic tool to evaluate the particle requirement is the Stokes' drag, in which the difference between the particle velocity and the fluid velocity can be estimated by

$$|V_p - U| = \frac{\rho_p d_p^2 \dot{V}_p}{18 \rho \nu} \quad (3-11)$$

where V_p is the particle velocity, U is the fluid velocity, $\dot{V}_p$ represents the total time derivative of the particle velocity, ρ_p and d_p are the density and diameter of the seeding particle, and ρ and ν are the density and the kinematic viscosity of the fluid, respectively. In this experiment, if the silver coated seeding particle with diameter of $10\mu\text{m}$ was used in blood analogue, ρ_p / ρ is approximately equal to 1, ν equal to 3.6 cS, the calculation shows that particle will only slip 1~2 mm/s even with the acceleration rate of 10^3 m/s^2 . This velocity difference between seeding particle and fluid element can normally be neglected for the flow past MHVs.

5) Image Concentration

The concentration of particle image is also critical for good PIV measurement. For cross-correlation technique, the number of particles required in each interrogation region (N_I) is required between 5~10. The image-density with relationship to the number of particles per volume C , the thickness of the laser sheet (Δz), the coefficient of magnification (M) is defined as:.

$$N_I = C \Delta z \frac{\pi l_{image}^2}{4M^2} \quad (3-12)$$

N_I can be checked by observing the number of particle images inside an interrogation area. More particles need to be added in the working fluid if N_I is too small.

N_I is also an important consideration for selecting the interrogation size. Small interrogation window can tolerate much higher displacement gradients, but it is not always satisfied with the required image density especially when considering the in-plane and out-of-plane loss of image pairs. Larger interrogation size, on the other hand, is relatively easy for densely seeding, but the mean displacement inside such large interrogation area tends to ignore fine flow details. Therefore a compromise should be made between these two sides.

3.4 Phase Lock Method

The heart rate of a normal human being is around 75 beats per minute, which corresponds to 0.8s cycle time. In particular, the heart valve only takes less than 50 ms to open and close respectively. To fully resolve the MHV opening and closing process in this experiment, at least five images should be captured, which requires a PIV system with a minimum working frequency of 100 Hz. A higher working frequency is undoubtedly required to resolve the squeeze flow, which is believed to occur on the order of microsecond. Unfortunately, this is beyond the limitation of the current PIV technology, which only has maximum pulse repetition rate of 30 Hz and 30 Hz frame rate for CCD camera. In order to solve this problem, a precise phase lock method was developed, which included a photoelectric sensor and a time-delay program, shown as Fig.3.13. For this method, the leaflet movement was captured by a photoelectric sensor, whose output signal was gradually time delayed to lock every specific instant in the cardiac cycle. The time delayed signal was then applied to trigger PIV system at the specified instants for images capturing.

Fig.3.13 Schematic of phase lock method

3.4.1 Photoelectric Sensor

A photoelectric sensor (Yamatake-Honeywell Co., Ltd, HPX-NT) was used to capture the motion of the leaflet. The sensor offers the highest scanning power and highest resolution (1/80 million) available in the current marketplace.

The receiver in HPX-NT accepts the light diffused from the target surface and output a corresponding high or low voltage depending on the amount of received light, shown in Fig.3.14. Due to the tiny light reflection point on the leaflet peripheral curvature (1mm width), the transparent material, and complex components surrounding the leaflet (including the enclosed transparent box and tube, blood analog suspend with glass particles), the leaflet position sensing is difficult.

Fig.3.14 Operation theory of photoelectric sensor

In order to obtain a stable signal when capturing the leaflet movement, the wall of the test section was kept as thin as 1~2 mm to minimize the distance between the sensor and the target. Super silver paint was applied on the specific position of the valve leaflet to

improve the reflectivity of the object. The sensor system, including the sensor head, wire cord and fiber cable, etc, was firmly fixed in a rack that was free of any vibration. The sensor was three-dimensionally adjusted for signal detecting. An external load was appended with the sensor output end to match the voltage requirement for the subsequent connected Intel 82C54 Programmable Interval Counter/Timer. The photoelectric sensor's main specifications and operational settings are listed in table 3.4 and Appendix I respectively.

Tabel 3.4 Photoelectric sensor specifications

Scanning distance	Thru: 290mm
	Diffuse: 110mm
Supply Voltage	10V-30V DC (ripple 10% max)
Current Consumption	50mA max
Output Mode	Open collector of NPN or PNP transistor
Output	Load Current: 100mA max. (resistive load) Saturation voltage: 1V max., 30 VDC max.
Response Time	500 μ s max.
Off-delay(selectable)	40 $\pm$ 4ms

Fig.3.15 Illustration of sensor's positions 1 &2
(cross section of MHV valve plane)

The sensor's positions were selected as site 1 and 2, shown in Fig.3.15 respectively for full field and closure field observation. At position 1, the sensor was triggered as the

valve occluder started to open, while at position 2, the trigger worked during closing phase as to capture the occluder closing movement during this interval.

3.4.2 Time Delay Programming

A time delay program was developed in-house using Boland C++ language, which controlled the operation of Intel 82C54 Programmable Interval Counter/Timer (1600 Data Acquisition Board, Keithley Instruments, Inc, USA), as shown in Fig.3.16.

Fig.3.16 Programmable timer configuration

The Intel 82C54 Counter/Timer consists of three independent, 16-bit, pre-settable down counters and a control register. Counter 0 was selected to generate programmable time delays with internal 100 kHz clock frequency. Counter 0 was written in the control byte, and meanwhile loaded with the required time delay value. The time delay process was initiated when the signal from the photoelectric sensor went high, as in Fig.3.17, then the counter 0 began to count down. At the end of time out, as the counter reaches zero, a rising edge of TTL signal was created and used as external trigger to the synchronizer.

In general, one rising edge photoelectric sensor signal was created in one cardiac cycle when the leaflet passed the position where the sensor was fixed. The time delay program was started by this rising edge and after specified time delay, it output another rising edge for PIV triggering. PIV system was thus set to operate at the same phase for specified cycles. Then a new time delay value was input and the images at that phase were captured again continuously.

Fig.3.17 Schematic illustration of sensor signal and time delayed signal

For this time delay method, the minimum delayed time reached $10\mu\text{s}$ and maximum value was $256^2 \times 10\mu\text{s} = 0.65536\text{ s}$. The time delayed value was easily adjusted by a step of $10\mu\text{s}$ in the range of $10\mu\text{s}$ to 0.65536 s . The advantage of this method was its accuracy, reliability and ease of manipulation.

3.5 Refractive Index Matching

For most of the optical diagnostic techniques that are applied to fluid dynamics, including PIV, one common difficulty is the refraction of light passing through gas-solid or liquid-solid interfaces at model and/or test section walls (Budwig, 1994). Commonly there are three methods to solve this problem:

- 1) the arrangement of test section and model;
- 2) applying additives to the working fluid to match the refractive index;

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3) image processing.

For the first method, it was suggested that a box with flat walls (view box) was constructed around the model when the flow field went through a curved inside wall. And the view box was filled with quiescent matching fluid, which was also pumped through the model (Budwig, 1994). For the second methods, the additives often used included sodium iodide (Subramanian et al., 2000; Baldwin, 1994), ammonium thiocyanate (Bluestein et al., 2000) or sodium thiocynate (Liu et al., 2000). Appendix II lists some properties of model material and the matching fluids for reference. Two disadvantages exist for the second method: 1) the property change of the resulting working fluid; 2) toxicity and corrosion of some additives. For the third method of image processing, it was proposed that a polynomial transformation equation could be used to formulate the relation between the image and the object via a precision grid in the region of interest. The coefficients were obtained from a minimization algorithm with least square condition. Then the images could be transformed pixel by pixel in terms of the transformation functions (Brucker et al., 2001). Although this technique keeps the original chemical composition of the working fluid, its accuracy depends on the order of the transformation equations. For near curvature boundary flow where image has the largest optical deformation rate, image processing method is seldom applied due to its relative poor accuracy.

In this experiment, sodium iodide (NaI) was finally selected as additives to match the refractive index of the model material, acrylic perspex. According to the prescription from Baldwin et al. (1994), 79 percent saturated aqueous sodium iodide, 20 percent pure glycerol and 1 percent of distilled water by volume were mixed together to produce the fluid required. The kinematics viscosity, refractive index and density of the mixture

solution were measured respectively and listed in table 3.5. The property of the sodium iodide and glycerin used in this experiment were also provided in this table.

Table 3.5 Properties of NaI, glycerin and mixture solution

Sodium Iodide (GCE Lab Chemicals)		Data source
Density	3.7g/cm ³	Hilger Crystals Ltd, (1999)
Refractive index	1.8	
Water solubility at 273K	159g/100g water	
Toxicology	Skin and eye irritant,	Oxford University
Glycerin (SINO Chemical Co, 99.5%)		
Viscosity at 20°C	1490cP	Chemical Handbook
Density	1.25kg/L	
Refractive index	1.47	
Mixture Solution		Measuring Instrument
Kinematics viscosity	3.6 cS	Controlled rate rheometer, contraves low shear 40 (USA)
Refractive index	1.4910	ATAGO 3T (Japan)
Density	1.69 g/ml	Electrical scale, measuring cylinder

Fig.3.18 Visualization test for refractive index matching

The measured refractive index of the mixture, 1.4910, was very similar to that of Perspex (1.49), material of the test section. A visualization test was also conducted by inserting an evenly spaced grid in the test section, which was immersed in the mixture. No visible deformation was observed after matching of refractive index, shown in Fig.3.18.

3.6 Shear Stress Calculations For PIV Images

Because of the periodic nature of the flow field, PIV images were analyzed using ensemble average techniques, in which average information are obtained by considering all the samples at the same instant of various cycles. According to Tiederman et al. (1988), cyclic variations exist for pulsatile flows, which necessitates the cycle resolved analysis, i.e., each cycle is treated as an independent part of flow. Each instantaneous axial velocity u_i and transverse velocity v_i can be divided into mean value U , V and high frequency turbulence fluctuations u'_i , v'_i as

$$u_i = U + u'_i ; \quad v_i = V + v'_i, \quad (3-13)$$

where i is the number of the cycle. U , V are the velocity average of various cycles at the same instant, which is calculated as

$$U = \frac{1}{N} \sum_{i=1}^N u_i ; \quad V = \frac{1}{N} \sum_{i=1}^N v_i \quad (3-14)$$

where N is the total number of cycles.

For the PIV maps in the following chapters,

$$\text{speed} = \sqrt{U^2 + V^2} ; \text{Vorticity} = \frac{\frac{\partial V}{\partial x} - \frac{\partial U}{\partial y}}{2} ; \text{Strain Rate} = \frac{1}{2} \left(\frac{\partial V}{\partial x} + \frac{\partial U}{\partial y} \right) ; \quad (3-15)$$

$$\text{Where } \frac{dU}{dy} = \frac{1000 \times [U(j, k+1) - U(j, k-1)]}{2[y(j, k+1) - y(j, k-1)]} ;$$

$$\frac{dV}{dx} = \frac{1000 \times [V(j, k+1) - V(j, k-1)]}{2[x(j, k+1) - x(j, k-1)]}$$

j, k : indexes along X, Y mapping direction.

The flow disturbance introduced by the MHV implantation may cause undesired strong fluid dynamics effects on the blood components, leading to the hemolysis and thromembolism complications. Thus its is necessary to quantify the level and distribution of the fluid stress when investigating the flow properties of a prosthetic heart valve. Laminar shear stress induced by the velocity gradients is one of the stresses for blood cell damaging, which is measured by:

$$\text{Laminar Shear Stress} = \mu \left(\frac{\partial U}{\partial y} + \frac{\partial V}{\partial x} \right) \quad (3-16)$$

For flow disturbances occurring near an artificial device, Reynolds stress should be considered to evaluate the mechanical load on the fluid components. The Reynolds stress is normally much higher than the corresponding viscous stress (Tennekes and Lumley, 1972). In particular, the Reynolds shear stress (RSS) τ_{xy} shows a strong positive correlation with hemolysis and platelet lysis (Sallam and Hwang, 1984), which is calculated as

$$\tau_{xy} = -\rho \overline{u'v'} = -\rho \frac{1}{N} \sum_{i=1}^N (u_i - U)(v_i - V) \quad (3-17)$$

Correspondingly, the Reynolds normal axial stress τ_{xx} and Reynolds normal transverse stress τ_{yy} are computed as

$$\tau_{xx} = \rho \overline{u'^2} = \rho \frac{\sum_{i=1}^N (u_i - U)^2}{N}, \quad \tau_{yy} = \rho \overline{v'^2} = \rho \frac{\sum_{i=1}^N (v_i - V)^2}{N} \quad (3-18)$$

The RSS measurement is dependent on the orientation of the reference axis. Thus the RSS measured in a particular coordinate system may not be representative of the maximum turbulent shear stress experienced by a fluid particle passing through a

considered point (Baldwin et al., 1993). It has been shown that with two-dimensional velocity measurement, the maximum obtainable RSS (also known as Major RSS) can be computed as follows (Barbaro et al., 1997; Grigioni et al., 2000b):

$$\text{Major RSS} = \rho \sqrt{\frac{1}{4}(\overline{v'^2} - \overline{u'^2})^2 + (\overline{u'v'})^2} \quad (3-19)$$

According to Nagaard et al.(1990) and Baldwin et al.(1994), the value of major RSS is about half of maximum τ_{xx} . The direction of maximum τ_{xx} is along the main flow direction, while the major RSS is in 45° with the main flow direction. Therefore, τ_{xx} can be approximately inferred from major RSS. The magnitude of τ_{yy} at most locations is normally negligible comparing to τ_{xx} (Maymir et al., 1998). Thus Reynolds normal stresses were not separately discussed in this work, and only viscous shear stress and major RSS were computed at every time step.

3.7 PIV Experiment Procedures

3.7.1 Preparation Work

The whole test rig was properly assembled. The working fluid was made up with glycerin, water and sodium iodide, which had the similar kinematics viscosity as human blood and the same refractive index as the test section. Silver coated hollow glass sphere of 8~10 μ m diameter was selected as the seeding particle due to its good reflectivity and appropriate image size. Sufficiently large seeding particle concentration was required when the systemic magnification was increased for the measurement of the narrowing channel flow. The number of the particles in each interrogation region, by checking their images on the monitor, should be no less than 5.

The motor was turned on and the piston movement was controlled by the simulated

motion curve. The two resistive and two compliances elements in the test loop were adjusted until physiological ventricular and aortic pressure waveforms and flow rate were maintained. These physiological signals were monitored during the whole experiment process.

All the PIV components, including monitoring computer, synchronizer, laser, laser controller, microscopy and CCD camera were properly connected. The suitable lightsheet optics (including cylindrical and spherical lens) was selected. The laser was installed on a tripod and positioned at the focal point of cylindrical lens to obtain the thinnest lightsheet. The planar laser sheet, with the thickness around 0.1 mm, cut cross the transverse center of the MHV plane, i.e., streamwise middle cross section. The flow field around the transparent MHV model was thus illuminated, as shown in Fig.3.19.

Fig.3.19 Illustration of the laser plane

CCD camera was located above the test section. The focal plane of the Nikkor lens or microscope was tuned until it coincided with the illumination plane, where the sharpest particle image can be observed on the monitor. Different magnifications were set up in this experiment according to various flow field sizes. For the global flow field observation in MHV, the field of view (FOV) was adjusted by varying the lens' $f^\#$ until its dimension in the transverse direction was a slightly larger than the aorta outer

diameter, i.e., 30mm, whereas in the axial direction the dimension covered the transparent MHV structure and also 7 mm upstream and downstream region. For the squeeze flow measurement, the microscope zoom was gradually increased to observe the narrowing gap flow under continuously improved spatial resolution.

3.7.2 Phase Lock Arrangement

The triggering sensor was fixed at different positions for different magnifications. For the global flow field measurement, the sensor was put immediately adjacent to the test section's back side, where the leaflet resides upon its fully closure. Thus the time-delay mechanism was started when the leaflet began to depart from the wall (i.e., the occluder about to open). Such position selection for the triggering sensor can avoid blocking the laser light path or the image capturing by CCD, and also any unfavorable effect for the image quality. In the squeeze flow measurement, the sensor was put underneath the test section and started the time delay during the leaflet closing phase.

The position of the photoelectric sensor was precisely adjusted until it captured the periodic movement of the leaflet. The sensor was properly connected with external load, 82C40 programmable counter, control computer and the synchronizer. The sensor output signal and the time delayed signal were simultaneously measured by an oscilloscope to make sure that the circuit worked properly. Various time delay values were input to the program and the leading edge of the delayed TTL signal was used as external trigger for the synchronizer.

3.7.3 PIV Measurements

The experiment was conducted in an isolated room for safe operation of laser. The room temperature was controlled at 25°C, and the relative humidity was kept at 65%. To

eliminate the external light, the room was totally isolated from the outer light source and only the laser provided the illumination for the flow field. The laser light was separated from the control console with a black screen. For the full field observation of the transparent bi-leaflet, Micro-Nikkor lens combined with CCD camera was used to capture the flow phenomenon in one complete cardiac cycle. For the detailed study of closing process, microscope enhanced with CCD was adopted. A X-Y traveling table with step adjustment of 0.01 mm in the X and Y directions was coupled with the microscope so as to precisely locate the interested region within an order of tens of micrometer. The position and focusing of the Micro-Nikorr lens and microscope were carefully checked so that the desired image field and best image quality were obtained. PIV system was then trial run to check whether there were 5~10 particles in every interrogation area. PIV images were taken for 100~200 continuous cardiac cycles with every input delay time value.

3.7.4 PIV Image Processing

With the known size of leaflet thickness (1mm) in the image, Insight software can automatically calculate the dimension of the actual observation field and thus finish the velocity calibration. When processing the PIV image in Insight software, the leaflet occupied region, where no particle presents, was excluded from the processing area. Interrogation size could be chosen from 32×32 pixels, 64×64 pixels and 128×128 pixels depending on the seeding density in the flow field. The corresponding vector files, which were created after processing, were loaded into Tecplot software to present PIV data with velocity, strain rate and vorticity, etc.

The vector length in the PIV map is proportional to the local velocity magnitude (e.g. the square root of the sum of the squares of the vector components U and V) and scaled

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to the length of the reference vector. The vector's arrowhead represents its in-plane direction, with the tail positioned at the data point. Reference vectors are not uniformly scaled in all the maps. They are specific to each figure group and to best accentuate the flow structure.

Chapter 4 Global Flow Field Across MHV Model

The global flow field across the transparent bi-leaflet MHV in one complete cardiac cycle, including opening process, full open stage and closing process was investigated using PIV technique. Micro-Nikkor lens incorporated with CCD camera was applied in this study. Due to the opening and closing occluder structure, the flow field past MHV demonstrates apparent fluid-structure interactions, which leads to its unsteadiness, turbulence and inhomogeneity. Different flow characteristics, together with occluder dynamics were demonstrated in various stages of the cardiac cycle. The laminar and Reynolds stress at every time step were calculated and regions with possible hemolysis and/or thromboembolism occurrence were also identified.

4.1 Flow Field Image in PIV

In order to capture the flow field at various instants, the time delay duration was kept changing. Fig.4.1 shows the simultaneously recorded ventricular pressure and trigger signal with 75 ms time delayed. The falling edge of the trigger signal represents the starting point of time delay counting, instant time-delay starting to count. The rising edge is the instant after specified time delayed, i.e., the instant PIV system was triggered. For each instant in the opening process, about 100 heart cycles were captured to study both the occluder dynamics and the flow characteristics, while in the full open and closing process, 200 cycles data were collected at every instant.

A typical flow field image is shown as Fig.4.2, where high seeding density is demonstrated as tiny white dots. The flow field is divided into three channels: one is the central channel between the two occluders, and two are the side channels between the occluder and the housing wall. The X-Y axis is oriented respectively as parallel (axial direction) and orthogonal (transverse direction) in the cross section plane of the valve.

And their velocity components are denoted as U and V respectively.

Fig.4.1 Ventricular pressure Vs trigger signal

Fig.4.2 Typical PIV image of global observation field

The angular position α of each occluder at every instant, shown in Fig.4.2, was obtained by positioning one set of X & Y pixel co-ordinate on the leaflet edge. The average angular/linear velocity can be deduced with the known time interval and the angular difference.

The dimension of the field of view (FOV) was measured as 31.8×32.3 mm. The image was divided into interrogation region of 64×64 pixel, each corresponding to a displacement calculated by the cross-correlation algorithm. With 50% Nyquist overlap applied in the image processing, 30×30 velocity vectors were derived in the measurement area.

Strong laser glare is observed around the valve ring region and this is due to the specific ring structure (two symmetric and removable ones). This region will not be considered in the results discussions.

4.2 Occluder Dynamics

Due to the presence of the MHV model image inside PIV frame, the occluders dynamics can thus be studied by analyzing their positions at specific instant. For the opening process, at each specified instant, good repetitiveness of the leaflet angular position between various cycles was demonstrated. With continuously increasing positive pressure gradient across MHV, the flow velocity keeps rising in the opening process, until the flow field becomes highly turbulent around the full open phase. The situation is further complicated by the onset of backflow in the closing phase, which makes it difficult to study the occluder dynamics in this stage. Thus only the leaflet opening dynamics was investigated. The angular positions of the two occluders were simultaneously measured at the instants of time step 0ms, 5ms, 10ms, 16ms, 20ms and 30ms in the opening process. It is shown that for every instant the maximum position difference between 100 cycles is within 1 degree. One reason for this angular deviation may be attributed to the tolerance of pixel marking in the PIV image. To ensure the reliability, the consecutive time steps were such selected that the angular difference between them was at least 2 times larger than the tolerance, i.e. about $2 \sim 2.5$ degree. The

angular positions averaged from different cycles were calculated for the two occluders.

The occluder average angular velocity (ϖ) and tip linear velocity ($\bar{V}$) were derived for each specific instant as follows:

$$\varpi = \frac{\alpha_1 - \alpha_2}{\Delta t}; \bar{V} = \varpi \times R \quad (4-1)$$

where α_1, α_2 : average angular position at two consecutive instants;

Δt : time duration between the two instants

R : =10.434mm, distance between the hinge and the occluder upper tip

The results of the leaflet average values for α and $\bar{V}$ in the opening phase are shown in Fig.4.3~4.4. In Fig.4.3, for all the specified instants, the maximum difference of average α between the upper and lower occluders is less than 1° . Thus the MHV model is also characterized by good symmetrical motion of the two occluders in the opening process, besides repetitive position is observed between various cycles, as discussed earlier.

The tip velocities for the two occluders, shown in Fig.4.4, feature similar value at every instant due to their good symmetric angular position. It is found that the occluder velocity starts to decelerate from the initial open position. The largest leaflet velocity, around 0.25 m/s is found within 5ms after MHV opening, whereas the lowest leaflet velocity is observed around the instant of 30 ms when the occluder is nearly fully open. The tip velocity averaged within 0~30ms is estimated about 0.19 m/s.

For the closing process, although no effort is taken to track the occluder position at every instant, it is found that MHV model takes about 40~50ms to fully close. The average tip velocity is thus estimated to be in the order of 0.15 m/s, which is comparable to that in the opening process.

Fig.4.3 Upper and lower leaflet angular position in the opening process

Fig.4.4 Upper and lower leaflet tip velocity in the opening process

4.3 Global Flow Field

Together with the PIV flow maps, ten velocity profiles are extracted to better compare the field characteristics in different region. The central channel is presented in three profiles: 1, 2 and 3, and profile 4, 5, 6 for side channel, as in Fig.4.5. There are also 2

profiles upstream of the valve (7,8) and 2 downstream (9,10). Profile 1 and 4 is along the hinge axis and profile 3 and 6 along the tip of the leaflet. For the upstream profiles, one is located at the immediate inlet of the MHV structure, while the other is 4mm further upstream. Similarly, one of the downstream profiles is positioned at the outlet of the central channel and the other one is 4mm further downstream. Due to the continuous motion of the occluder, some profiles may not have fixed X co-ordinate.

Fig.4.5 Illustration of the positions for ten velocity profiles

4.3.1 Opening Process

The MHV model takes about 35 ms to travel to its full open position. The flow maps, velocity profiles and stress distribution at the time step of 0ms, 5ms, 10ms, 16ms, 20ms and 30ms are shown in Fig.4.6~Fig.4.26.

At $t=0$ as in Fig.4.6, the occluders are fixed at the fully closed position, by the downstream backflow acting on them. Meanwhile, the forward flow already starts to develop in the upstream region. Both the backflow and forward flow are weak with the magnitude around 0.1 m/s.

At the instant of $t=5$ ms as in Fig.4.7, the two occluders are displaced about 7° from the fully closed position. Three flow passages thus emerge across the MHV, i.e., one

through the central channel and two symmetrically across the side channels. The three channel flows are observed with the peak velocity of 0.52 m/s at the central inlet and two outlets of the side channels. The maximum Reynolds number (Re) at this instant is calculated to be around 3611 (aorta diameter of 25mm as the characteristic dimension). The side channel flow follows the occluder's surface and pushes the MHV to open. In turn, leaflet's motion drives the fluid, resulting in the flow perpendicular to the occluder surface. Away from this surface, the flow field in the central channel is the combined reactions of central jet spread and occluder driving effect, leading to the flow parallel to the axial direction. The central jet separates from the occluder surface immediately downstream of the entrance nozzle. Downstream of the hinge area, the jet re-attaches to the occluder surface due to the occluder rotation and central jet spread. The spread of the central jet between the two occluders may offer obstruction for the MHV opening motion. Flow separation is also demonstrated at the occluder ending tips. The largest vorticity is found at the hinge area and the occluder tips from the vorticity mapping, suggesting the maximum intensity of vortex at these regions.

For the velocity profiles inside the central channel as shown in Fig.4.8(a), the jet strength is dramatically reduced from profile 1 to profile 2. Profile 3 has a larger velocity magnitude than that along profile 2. This is attributed to the higher occluder velocity at the tip and thus stronger pushing effect to the nearby fluid. The flow velocity at the occluder tip is about 0.2 m/s, which is comparable to the average occluder speed of 0.25 m/s during the 5ms interval, as in Fig.4.4. For the profiles within the side channel, in Fig.4.8(b), the flow speed is continuously increasing along the forward flow direction, which results from the gradually narrowing of the passage's cross-section area. For the velocity profiles in the upstream as in Fig.4.9(a), profile 7 basically maintains a channel flow profile. Whereas for profile 8, immediately upstream of the bi-

leaflet, three distinct channel flow profiles with larger velocity gradient are observed, which is produced by the presence of two occluders structure. For the downstream profiles in Fig.4.9(b), the velocity magnitude issuing from side channel is larger than that from the central channel. More closer to occluder structure, the larger velocity gradient is found between the downstream flow of the central channel and side channel. From the stresses distributions shown in Fig.4.10, it is found that immediately downstream of the central nozzle, the jet separates from the occluder structure, causing large velocity gradient near the hinge area. Large fluid deformation is also demonstrated at the occluder ending tips due to the flow separation. Relative higher viscous shear stress is observed at these two regions. The strongest momentum mixing is confined to the hinge area. The maximum major RSS reaches 72 N/m^2 in magnitude.

Fig.4.6 Velocity and vorticity maps at t=0

Fig.4.7 Velocity and vorticity maps at t=5 ms

Fig.4.8 Velocity profiles inside (a) central channel (b) side channel at $t=5$ ms

Fig.4.9 Velocity profiles in the (a) upstream (b) downstream at $t=5$ ms

Fig.4.10 Stresses distributions at t=5ms

At the instant of $t=10\text{ms}$ as in Fig.4.11, the occluder continues to open to the angular position $\alpha = 37^\circ$. All the three channel flows become stronger with maximum flow velocity increasing to 0.55 m/s . The maximum Re is 3820 at this instant. The peak vorticity occurring at hinge area and the occluder tip velocity increases its magnitude too.

For the profiles inside the central channel, as in Fig.4.12(a), the central jet becomes stronger with its width expanded, arising from the increased flow speed and enlarged central nozzle. The other two central channel profiles, profile 2 & 3 demonstrate reduced flow velocity adjacent to the occluder downside. The occluder tip velocity is 0.2 m/s at this instant, as in Fig.4.4, whereas the flow velocity at the occluder downside tip is also around 0.2 m/s (profile 3). It is suggested that the occluder rotation is more affected by the resisting action of the central jet spread than the pushing effect of the side channel flow. All the three profiles in the side channel and two profiles in the upstream and downstream region, as in Fig. 4.12(b), Fig.4.13(a) and (b), show increased velocity in magnitude. Profile 6, at the exit of the side channel is observed with flow separation since larger velocity gradient occurs around the occluder tip. The immediate upstream profile 8 and downstream profile 9 also demonstrate more dramatic velocity gradient around the junction area of the occluder structure.

Due to the stronger central jet and flow separation immediately downstream of the channel inlet, higher levels of turbulent mixing is produced around the hinge area. Increased major RSS stress is found here, as shown in Fig.4.14. The flow separation at the hinge also causes stagnation zone adjacent to the occluder surface, which is prone to thromboembolism complication. The maximum viscous shear stress also increases at the occluder end tips and hinge region, where the increased velocity gradient produces elevated shear layer.

Fig.4.11 Velocity and vorticity maps at $t=10$ ms

Fig.4.12 Velocity profiles inside (a) central channel (b) side channel at $t=10$ ms

Fig.4.13 Velocity profiles inside the (a) upstream (b) downstream at $t=10$ ms

Fig.4.14 Stresses distributions at t=10ms

At the instant of $t=16\text{ms}$ as in Fig.4.15, flow velocities in the three channels further increase to around 0.7 m/s , with maximum Re up to 4860. The vorticity strength at the hinge and occluder tip also becomes stronger. For the velocity profiles inside the central channel, as in Fig.4.16(a), the maximum magnitude of profile 1 is further increased and jet width also expands. Contrary to the previous instants, profile 2 starts to have larger velocity than profile 3 due to the stronger spread of the central jet momentum. From profile 3, the flow velocity immediately downstream to the occluder tip, is now reduced to 0.15 m/s , whereas the MHV average tip velocity is 0.147 m/s during this interval, in Fig.4.4. All the profiles in the side channel, upstream and downstream regions, as shown in Fig.4.16 (b), 4.17(a) &(b), are found with increased velocity. Profile 6 demonstrates larger velocity gradient close to the occluder tip, which is attributed to the more apparent flow separation and wake formation. The upstream profile 7 and 8 demonstrate higher velocity at the channel core, implicating more evidence of the central channel flow.

For the stress distribution at this instant, as shown in Fig.4.18, the region where the maximum major RSS occurs extending about 2 mm downstream, causing a relatively elongated region of elevated stress. Its magnitude is increased to 90 N/m^2 , arising from the more intense turbulent momentum transfer at this instant. The viscous shear stress also increases with the magnitude reaching 1.7 N/m^2 .

Fig.4.15 Velocity and vorticity maps at t=16 ms

Fig.4.16 Velocity profiles inside (a) central channel (b) side channel at $t=16$ ms

Fig.4.17 Velocity profiles inside the (a) upstream (b) downstream at $t=16$ ms

Fig.4.18 Stresses distributions at t=16 ms

At the instant $t=20\text{ms}$ and $t=30\text{ ms}$, as shown in Fig.4.19 and Fig.4.23, the flow momentum inside the three channels is continuously increasing with the peak velocity reaching 0.79 m/s and 1.19 m/s respectively. The corresponding maximum Re is 5486 and 8264. The vorticity becomes further stronger at the occluder tips and hinge region. Comparing to that at $t=20\text{ ms}$ in Fig.4.20(a), profile 1 at $t=30\text{ ms}$ has its strength and width increased, as in Fig.4.24 (a). Profile 2 at both instants show larger velocity gradient than profile 3, suggesting that the central jet weakens quickly with the distance. All the other profiles in the side channel, upstream and downstream regions, as shown in Fig. 4.20(b), Fig.4.21, Fig.4.24(b) and Fig.4.25, have their velocity continuously raised in magnitude. The wake at the occluder tip becomes stronger, as shown from profile 6. The immediate upstream profile 8 and downstream profile 9 are observed with increasing velocity gradient between the central channel and side channel.

The stresses distributions maps, as in Fig.4.22 and Fig.4.26, are shown with continuously elevated shear stress and major RSS. Around the MHV hinge, strong flow separation causes large velocity gradient occurring inside a more elongated region. Both the size and intensity of the wake at the occluder tip is increasing. The abovementioned regions thus produce the maximum viscous shear stress and turbulent shear stress.

Fig.4.19 Velocity and vorticity maps at $t=20$ ms

Fig.4.20 Velocity profiles inside (a) central channel (b) side channel at $t=20$ ms

Fig.4.21 Velocity profiles inside the (a) upstream (b) downstream at $t=20$ ms

Fig.4.22 Stresses distributions at t=20ms

Fig.4.23 Velocity and vorticity maps at t=30 ms

Fig.4.24 Velocity profiles inside (a) central channel (b) side channel at $t=30$ ms

Fig.4.25 Velocity profiles inside the (a) upstream (b) downstream at $t=30$ ms

Fig.4.26 Stresses distributions at t=30ms

4.3.2 Fully Opened Stage

The flow field across MHV continue to accelerate after $t=30\text{ms}$ till it reaches peak systole at $t=50\text{ms}$. Afterwards, the flow field experiences deceleration phase, while keeping fully opened position unchanged for about 240ms. Typical instants of 50ms, 220ms, 270ms were selected to illustrate the evolutions of the flow field during this interval, as shown in Fig. 4.27~4.38.

At the instant of 50ms, as in Fig.4.27, the flow velocity in the central channel can be as large as 1.7 m/s and the maximum Re now reaches nearly 11800. Slight occluder fluttering is observed during peak systole phase. It may be attributed to the pressure fluctuation in the wake of the occluder, whose separation causes an unstable opening behavior and flutter of that occluder. Strong vorticity is demonstrated at the immediate downstream of the MHV and hinge area, where the strong flow separation occurs. For the central channel profiles as in Fig.4.28(a), the central jet intensity is further strengthened, and the maximum magnitudes at profile 2 and profile 3 are similar. The profile 6, in Fig.4.28(b), is observed with larger velocity gradient, suggesting the increasingly strong wake at the occluder tip. The immediate downstream profile 9, in Fig.4.29(b), demonstrates “three finger” shape, while further downstream, as in profile 10, the central flow weakens faster than that from the side channel.

The maximum shear stress is still found around the hinge region, as shown in Fig.4.30, where large local deformation and intense turbulent momentum mixing occur. Large stress is also identified inside the central channel, resulting from the strong central jet penetrating downstream of the hinge and its separation from the occluder surface. The occluder tip is observed with elongated wake, extending further downstream of the MHV structure. The wake is shown with large vorticity and shear layer.

Fig.4.27 Velocity and vorticity maps at t=50ms

Fig.4.28 Velocity profiles inside (a) central channel (b) side channel at $t=50$ ms

Fig.4.29 Velocity profiles inside the (a) upstream (b) downstream at $t=50$ ms

Fig.4.30 Stresses Distributions at t=50ms

At the instant of $t=220\text{ms}$ and $t=270\text{ms}$, shown in Fig.4.31~4.38, the flow in the three channels becomes continuously weakened. The peak velocity decreases to 0.78 m/s at $t=220\text{ms}$, and further decrease to 0.26 m/s 50ms later. The maximum Re also decreases to about 5417 and 1806 correspondingly. Due to the existing weaken central jet, the maximum velocity magnitude of profile 1 at $t=220\text{ms}$, is still larger than other two profiles, as in Fig.4.32(a). However at $t=270\text{ms}$, the central jet flow almost disappears, profile 3 in turn is found to have the largest magnitude as in Fig.5.37(a). The strengths of the profile 1 and 2 at $t=270\text{ ms}$ thus rapidly weaken. Similarly, the immediate upstream profile 8 in Fig.4.33(a) has three peak velocity, whereas at $t=270\text{ms}$, in Fig.4.37(a), profile 8 is shown with comparatively lower magnitude for the central channel. As for the side channel profiles, as shown in Fig.4.32(b) and Fig.4.36(b), the velocity magnitude and the wake strength is continuously weaken with time elapse. The downstream profiles at both instants, shown in Fig.4.33(b) and Fig.4.37(b), demonstrate relative stronger side channel flow than the central channel, although their magnitude are both reduced. With the decelerating of the forward flow momentum, the vorticity also decreases during this interval.

The peak viscous shear stress and major RSS is continuously reducing after peak systole. At the instant of 220ms , the region for the maximum stresses is still located around the hinge and occluder downstream wake, as shown in Fig.4.34. With the central jet disappearing at $t=270\text{ms}$ as in Fig.4.38, the downstream wake and central channel demonstrates relative higher shear stresses instead of the hinge area.

Fig.4.31 Velocity and vorticity maps at t=220ms

Fig.4.32 Velocity profiles inside (a) central channel (b) side channel at $t=220$ ms

Fig.4.33 Velocity profiles inside the (a) upstream (b) downstream at $t=220$ ms

Fig.4.34 Stresses distributions at t=220ms

Fig.4.35 Velocity and vorticity maps at t=270ms

Fig.4.36 Velocity profiles inside (a) central channel (b) side channel at $t=270$ ms

Fig.4.37 Velocity profiles inside the (a) upstream (b) downstream at $t=270$ ms

Fig.4.38 Stresses distributions at t=270ms

4.3.3 Closing Process

After peak systole, forward momentum through the three channels continuously weakens. The closing phase starts at the onset of the back flow, which pushes the occluder from the fully opened position to close. Typical velocity maps and stresses distributions at the instant of 310 ms, 330 ms, and 350 ms, as shown in Fig.4.39~4.44 are selected to demonstrate the evolution of this process.

At $t=310\text{ms}$, as shown in Fig.4.39, the back flow initiates in the central channel, which separates into three streams. One stream reverses back through the central orifice with peak velocity of 0.2 m/s. The other two regurgitant streams, perpendicular to the occluders' surfaces, push the leaflets to close and launches the backflow upstream of the occluder trailing tip. The asymmetry of these two jets may result in the asymmetrical closing movement of the two leaflets. Downstream of the occluder trailing tip, forward flow is still observed with velocity of 0.11 m/s. These results are similar to those from Subramanian et al. (2000).

For the stresses distributions at this instant, as shown in Fig.4.40, larger shear stress is produced inside the central channel, where the uneven distributed back flow causes relative large velocity gradient.

With the closing phase proceeds further, at $t=330\text{ms}$, as shown in Fig. 4.41, the regurgitant flow becomes more prominent. Due to the stronger backflow in the central channel and reduction of the nozzle cross-section area, the peak velocity reaches 0.36 m/s inside the central nozzle. The large backflow around the nozzle may help to wash-out the thrombi accumulated at the central jet boundaries in the opening process. The stronger backflow also accelerates the occluders closing motion, which in turn strengthens the back flow upstream of the occluder trailing tips. Downstream of the trailing tip, the forward flow tendency further weakens.

Both the peak laminar shear stress and major RSS increase at this instant, as shown in Fig.4.42. There is a strong reverse jet flow that rushes out from the nozzle of the central channel, which encounters with the ambient low flow in the upstream region, causing higher shear layer.

When $t=350\text{ms}$, as shown in Fig.4.43, the gap between the leaflet trailing tip and the housing wall decreases to about 1~2 mm. The regurgitant peak flow velocity further increases to 0.49 m/s in the central nozzle and about 0.3 m/s upstream of the occluder trailing tip.

The stress at this instant, as shown in Fig.4.44, also continuously increases, with the maximum magnitude of 43 N/m^2 still occurring at the nozzle exit. The stronger nozzle jet results in larger velocity gradient when it mixes into the wide upstream region.

The vorticity keeps increasing with the leaflet approaching the housing wall. Its peak magnitude is normally found around the leaflet trailing tip and exit of the central channel nozzle.

Fig.4.39 Velocity and vorticity maps at t=310ms

Fig.4.40 Stresses distributions at t=310ms

Fig.4.41 Velocity and vorticity maps at t=330ms

Fig.4.42 Stresses distributions at t=330ms

Fig.4.43 Velocity and vorticity maps at t=350ms

Fig.4.44 Stresses distributions at t=350ms

4.4 Discussions

In this work, the flow characteristic and stresses distributions inside the occluder structure were acquired from the PIV investigation of a transparent MHV model. The PIV system triggering was sampled from the displacement of the occluder position that enabled the simultaneous measurement of the occluder motion and the flow field across it. The ensemble-average technique was applied to calculate the average velocity and turbulent shear stress at each instant in the opening process, fully opened stage and closing process, so that regions with possible thrombus formation and/or blood damage can be identified.

The occluder dynamics in the opening process is affected by the continuously changing flow field, while MHV rotation also influences the flow evolution. At every specified instant, MHV demonstrates good symmetrical motion of the two occluders and also good repetitiveness among various cycles. The occluder motion is found decelerating in its opening process due to the more prominent resisting effect of the central jet expansion.

Three high-speed channel flows are exhibited through the MHV model and their magnitudes increase till peak systole. After systole, the forward flow momentum decelerates until the backflow initiates in the central channel. The back flow pushes the occluder to close and the regurgitant flow becomes stronger with the acceleration of the occluder closing process. Table 4.1 summarizes the related values in the opening process, fully opened stage and closing process.

Before peak systole, the magnitude of the viscous shear stress and major RSS is found to increase continuously. These peak values basically locate around the hinge area and the occluder trailing wake. The strong central jet is shown with separation from the occluder surface at the hinge, causing large velocity gradient and intense turbulent

momentum exchange. Moreover, the flow separation at the jet boundary also causes relative stagnant flow field adjacent to the surface. With the central jet becoming stronger and extending downstream inside the central channel, the peak stress becomes larger in magnitude and its region also extends about 2 mm downstream of the hinge. The occluder trailing tip is observed with wake formation, whose intensity and dimension both are increasing till peak systole. The elevated shear layer thus occurs, resulting in large local deformation and turbulent mixing.

Table 4.1 Max. magnitudes of various flow parameters at different instants

	Max.speed (m/s)	Re	Max. vorticity (/s)	Max. viscous shear stress (N/m ²)	Max. major RSS (N/m ²)
5ms	0.52	3610	88	1.21	71.98
10ms	0.55	3820	91	1.30	77.40
16ms	0.70	4860	121	1.70	90.73
20ms	0.79	5486	149	1.87	115.14
30ms	1.19	8264	190	2.48	270.69
50ms peak systole	1.7	11800	243	2.51	501.16
220ms	0.79	5417	126	2.01	159.7
270ms	0.26	1806	27	0.57	14.98
310ms	0.2	1389	37	0.31	8.6
330ms	0.36	2500	79	0.68	41.1
350ms	0.49	3403	112	0.79	42.8

The downstream MHV flow field in the opening process under the pulsatile flow condition has been widely investigated (Brucker 1997; Lim et al., 2001; Subramanian et al., 2000; Brucker et al., 2002). They showed that the maximum velocity downstream of the MHV is within 1.5~2 m/s at peak systole which is comparable to the current study.

The finger shape profile at peak systole reported for SJM, Sorin and Edward Tekna bi-leaflet MHV (Barbaro et al., 1997) is also demonstrated in this study at the immediate downstream of the occluder. The reported maximum turbulent shear stresses in the MHV downstream region differ in various studies, from tens of N/m^2 to several hundred N/m^2 due to the different flow regimes and valves used (Yoganathan et al., 1986; Barbaro, et al., 1997; Lim et al., 2001; Liu et al., 2001; Balducci et al., 2004). For the first time, the current work provides the stress distributions inside the MHV structure. It is shown that the maximum value is normally located at the wake of the occluder tip, and especially around the hinge area, where the flow separation causing the strongest turbulent mixing. At the peak systole, the major RSS reaches 500 N/m^2 . This is larger than the threshold for hemolysis reported by Sallam and Hwang (1984). Moreover, the stagnation zone arising from the flow separation at the central jet boundary may cause thrombus formation. Possible ways to reduce the shear stress around the hinge region is to improve the occluder leading tip and occluder opening dynamics, so that the flow separation would be delayed downstream, where large velocity gradient can be decreased if the central jet spread would gradually re-attach to the opening rotation occluder.

The peak shear stress region is extended downstream around peak systole instant, possibly resulting from the relative small occluder opening angle. The larger fully opening angle, the less likely occurrence of the abovementioned phenomenon. However, large opening angle may also cause stronger occluder motion upon its fully closed, which may result in higher shear stress and also an increase in the risk of cavitation occurrence. The determination of the MHV opening angle should be compromised between the two sides.

After peak systole, the maximum stress magnitude decreases due to the reduced intensity of flow in the three channels. The peak stress region is no longer located around the hinge area as the central jet disappears. Instead it moves to the middle section of the central channel, where the flow is relatively inhomogeneous.

Although the shear stress is also increasing with the occluder gradually closing, its magnitude is generally less than during the opening process, possibly due to the relatively weakened back flow. A relative strong reverse jet flow is found rushing out from the nozzle of the central channel. Higher shear layer is produced when this jet encounters with the ambient low velocity flow in the upstream. Large shear stress is demonstrated around this region with peak RSS reaching 43 N/m^2 upon MHV nearly full closed.

In general, the flow field around MHV is unsteady, turbulent and inhomogeneous. The central and two side channel flows keep changing in one cardiac cycle, and they interactively affect the occluder dynamics. In turn the occluder motion also influences the two channels' flow. These experimental results provide good validations for future CFD study of the fluid dynamics across prosthetic heart valve coupled with the simulation of the fluid-structure interaction.

It should be noted that the transparent MHV was used in this work. The density is slightly lighter lower than that of the commercial MHVs. Hence the opening & closing dynamics and the resultant local flow field may be affected.

Chapter 5 Squeeze Flow Field Observation

MHV closing process, especially around full closure, including squeeze flow, rebound and leakage jet attracts much interests in recent years. Cavitation potential and higher turbulent shear stress have significant link to the abovementioned phenomena (Bluestein et al., 1994; Wu 1996; Maymir et al., 1997; Meyer et al., 1997; Manning et al., 2003; Lim et al., 2003).

In this study, a complete transparent MHV model enables the investigation of channel flow between the occluder and the orifice wall. With the closing process proceeding, the channel becomes increasingly narrower until zero gap when the valve is considered fully closed. The gap effect after MHV full closure and the corresponding leakage flow are not studied in this work.

Fig.5.1 Observation field with (a) micro-Nikkor lens (b) microscope

A Stemi 2000-C microscope was used to observe the magnified channel flow as the occluder tip approaches the orifice wall, as shown in Fig.5.1(b). Gap 1 and Gap 2 are the distances between the wall and the leaflet leading tip and the trailing tip respectively. Three profiles were selected within this narrow gap, i.e., profile 1 along the leaflet leading tip, profile 3 along the trailing tip and profile 2 between profile 1 & 3. Velocity profiles

including the speed and its U, V component were extracted along the abovementioned three profiles and plotted against Y-axis.

5.1 Leaflet Closure Velocity

In order to facilitate capturing of the MHV closure phenomenon, the starting point for the time delay was sampled during the leaflet closing movement. The interval Δt , which was between zero instant to the leaflet fully closure instant, was obtained. Due to the variation between beat to beat, Δt was selected as the duration at which the occluder reaches its full closure position over 85% of 100 continuous hearts cycles.

The MHV average tip velocity within Δt , can therefore be calculated by analyzing leaflet position in the PIV image. By finely tuning the sensor position, the zero instant can be started from different positions in the MHV closure process. With the magnification increasing, the zero instant was selected gradually closer to the leaflet fully closure position. Thus the leaflet tip velocities within the decreasing gap dimensions were obtained, shown as table 5.1. In this table, α , is the angle the leaflet oriented with the transverse direction, shown as Fig. 5.1(a). It demonstrates that the leaflet tip velocity accelerates with the closing process proceeding.

Table 5.1 Average leaflet velocity within various gap dimensions

Magnification (M)	Gap 1 at t=0 ms (mm)	α at t=0ms	α at fully closure	Δt (ms)	Average Leaflet tip velocity (m/s)
0.28	5.2	72°	30°	55	0.14
1.8	4.24	63°	30°	35	0.16
2.6---3.2	0.98	40°	30°	6.5	0.24

6.2 Flow Field Results

1) $M=1.28$

Under this magnification, the field of view is about $4.8 \times 4.9 \text{ mm}$ and zero instant was selected at the gap distance around 4.2 mm . One example of simultaneously recorded time delayed signal and ventricle pressure is shown as Fig.5.2.

Fig.5.2 Ventricle pressure Vs time delay

Series of results at various gap dimensions are shown in Fig.5.3~5.6. The velocity map, strain rate and vorticity distributions are presented in the left column, while the speed and its U, V component profiles are illustrated in the right column.

When the gap is about $1.15 \sim 1.4 \text{ mm}$, as shown in Fig.5.3(a), the leaflet is pushed to close by the fluid in the backwind region with flow velocity perpendicular to the surface. In the upwind region, the fluid adjacent to the leaflet is driven in motion by the leaflet rotation also with flow velocity perpendicular to the solid surface. While apart from the surface, the flow field is the combined result of the back flow through the gap and the occluder driven flow, which results in the flow velocity diverting to be parallel to the axial direction. The

peak velocity is found to have a magnitude of 0.27 m/s. A vortex flow is formed around the leaflet trailing tip, resulting from the solid-liquid interaction at the tip and the backflow around the gap. The maximum vorticity are found at this location, with a magnitude of 182 /s.

The speed and its U, V component profiles inside the gap, as shown in Fig.5.3(b), are shown with the largest magnitude along the leading tip, while those at the trailing tip are the minimum.

When the gap decreases to about 0.65~0.95 mm as in Fig.5.4(a), the backflow inside the gap strengthens. Both the backflow momentum and the solid-liquid interaction around the occluder become stronger as the peak flow velocity reaches 0.36 m/s. The intensity of the vortex at the trailing tip increases to 300 /s and the peak strain rate also rises to 233 /s. Meanwhile, the magnitudes of the speed and its U component profiles increase, as in Fig.5.4(b). The magnitude of the V component shows negligible change comparing to the previous instant.

Fig. 5.3 (a) velocity map, strain rate and vorticity distributions (b) velocity profiles when gap 1.15-1.4 mm

Fig. 5.4 (a) velocity map, strain rate and vorticity distributions (b) velocity profiles when gap 0.65-0.95 mm

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When the gap continuously decreases to about 0.5~0.65 mm, as shown in Fig.5.5(a), the backflow momentum inside the gap further strengthens resulting from the increased aortic-ventricular pressure gradient and ever narrowing gap dimension. The resultant flow in the upwind region, which combines the backflow through the gap and the structure driving flow, is stronger than that in the backwind region. The maximum velocity at the leading tip increases to 0.4 m/s. The peak strain rate and vorticity also continuously increase.

The speed and its U component profiles, as shown in Fig.5.5(b), also have their magnitude further increase.

Fig. 5.5 (a) velocity map, strain rate and vorticity distributions (b) velocity profiles when gap between 0.5-0.65 mm

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2) $M=3.2$

In order to further improve the spatial resolution inside the gap, the magnification was increased to 3.2 and the field of view now decreases to about $2.70 \text{ mm} \times 2.73 \text{ mm}$.

Fig.5.6(a) shows the flow field when the gap is about $0.21 \sim 0.35 \text{ mm}$. The stronger backflow momentum inside the gap and in the backwind region, together with the acceleration of the occluder motion, results in the maximum velocity of 0.5 m/s at the upwind region. The vorticity around the trailing tip reaches about 800 /s . The inside gap profiles for the speed and its U component, as shown in Fig.5.6(b), also increase in magnitude with the peak value up to 0.4 m/s at the leading tip.

With the gap further decreases to about $0.19 \sim 0.3 \text{ mm}$ as in Fig.5.7(a), the maximum flow velocity in the leaflet immediate upwind reaches 0.8 m/s . The magnitudes for the three speed profiles further increases, as shown in Fig.5.7(b), with the one at the leading tip up to 0.7 m/s .

When the gap is less than 0.1 mm , as illustrated in Fig.5.8(a), the jet with magnitude of 2.3 m/s is observed at the exit of gap. From zoomed in map as in Fig.5.8(b), it was shown the high speed flow coming out from the leading tip encounters a sudden geometry expansion and mixes with relative weak flow in the upwind region, which may result in local pressure reduction for cavitation occurrence. The intensity of the vortex around the leading tip reaches about 2400 /s .

Fig.5.6 (a) velocity map, strain rate and vorticity distributions (b) velocity profiles when gap between 0.21-0.35 mm

Fig. 5.7 (a) velocity map, strain rate and vorticity distributions (b) velocity profiles when gap between 0.19-0.3mm

Fig.5.8 (a) velocity map, strain rate and vorticity distributions (b) zoom in map when gap $< 0.1\text{mm}$

5.3 Discussions

The occluder closure dynamics and narrowing channel flow were investigated using DPIV technique incorporated with a microscope. For the first time, this study experimentally illustrates in detail the gradual development of “squeeze flow” phenomenon.

The average occluder tip velocity is found to increase as the MHV gradually approaches the orifice wall. The average tip velocity is shown to have the magnitude of 0.24 m/s within the final 10°. It was suggested that the occluder closing motion has strong correlation with the valve type, heart rate and cardiac output, etc (Chandran et al., 1994; Wu et al., 1994; Wu et al., 1995a; Wu et al., 1995b). Wu (1995a) obtained the mitral MH valve impact velocity of 1.5~2 m/s for its last 3° excursion using laser sweeping technique (LST). Similarly it was also shown that the value for bileaflet MHVs ranging from $0.52 \pm 0.15 \sim 1.64 \pm 0.38$ m/s (Wu et al., 1995b). Although the PIV method may lack the high resolution of LST in measuring the occluder micro motion, it features the advantages of fast and convenience, together with the simultaneous observation of the nearby flow field.

The flow around the narrowing channel is demonstrated with the gap width continuously decreasing from 1.15~1.4 mm to 0.19~0.3 mm, and till less than 0.1mm. The gap spatial resolution is improved by increasing the system magnification. When the occluder gradually approaches the orifice wall, the backflow momentum increases both in the backwind region and inside the gap. The upwind region flow is the combined result of the occluder-driven and the ever narrowing gap flow, which produces larger velocity than that in the backwind region. The peak strain rate is normally found along the leaflet surface and tips, where the relative larger velocity gradient is formed. The maximum vorticity exists around the occluder trailing tip and its intensity becomes stronger with the leaflet

approaching the orifice wall. Table 5.2 summarizes the peak velocities, strain rate and vorticity under various gap dimensions.

Table 5.2 Measurement results at different gap dimensions

Gap (mm)	Max.flow velocity upstream (m/s)	Max.flow velocity downstream (m/s)	Max. strain rate (1/s)	Max.vorticity (1/s)
1.15~1.4	0.26	0.26	175	182
0.65~0.95	0.36	0.36	233	300
0.5~0.65	0.40	0.38	406	418
0.21~0.35	0.51	0.42	810	793
0.19~0.3	0.83	0.55	820	932
<0.1	2.3		2266	2442

Due to the strong occluder driven at the leading tip, the maximum fluid speed and its U component are always observed at profile 1 with the gap continuously narrowing. The profile 3 is basically shown with the minimum magnitude, where the wake is formed and gap dimension is relative larger. V component, on the other hand, shows decreasing magnitude when the gap becomes narrower. When the gap decreases to less than about 0.3 mm, profile 1 basically has zero V component due to the more prominent parallel tip velocity from the occluder driven effect. V component along profile 2 & 3 also reduces its magnitude with the gap dimension continuously decreasing.

With the gap further decreasing to less than 0.1 mm, the high speed flow of about 1.7-3m/s is observed at the exit of the gap. The high speed jet, when rushing out the gap, mixes with the relative low speed flow in the upwind region and causes large vorticity of up to 2442/s.

The possible low pressure center may thus be produced for potential cavitation inception to occur. Meanwhile, the large strain rate is also found outside the gap due to the strong velocity gradient. This may characterize increased shear damage to the formed elements of blood.

The squeeze flow is a localized, rapidly evolving transient phenomenon, which are affected by the geometry of the gap and the closing leaflet velocity (Hwang,1998). Bluestein (1994) found in his CFD computation that the squeezing flow velocity for a mitral MHV can be as high as 30 m/s. In a test rig simulating single closing event, Chandran (1998a) roughly estimated the squeeze flow of 12 m/s by capturing the cavitation bubble motion. LDA measurement of a Bjork-Shiley MHV detected velocity spike around 3 m/s at 1mm away from the valve. It was hypothesized to be the evidence of squeeze flow (Maymir et al.,1998). This experimental investigation provides a direct measurement of “squeeze flow” to support these finds.

The difference between the current PIV measurement with other works possibly results from different valves and experiment set ups. The one shot test rig produces higher pressure gradient upon closure, while the monostrut MHV tends to close faster than bileaflet MHV.

Chapter 6 Transient Pressure and Closure Flow Measurements

The cavitation in MHV is influenced by many factors, including pressure loading rate, occluder material, valve holder flexibility, valve size and working fluid properties, etc. Among them, the valve design factor attracts particular interests, since it is found that some valves with specific features are more vulnerable to cavitation inception (Bluestein et al., 1994a; Lee et al., 1996a; Chandran et al., 1997). In order to study the influence of valve design for the MHV closure flow field and potential cavitation occurrence, two additional transparent bi-leaflet MHV models with different tip angles were fabricated. The transient pressure at the two ends of the gap was simultaneously measured by two micro-tip pressure catheters around the occluder's closure instant. The closure flow fields for each valve model were also investigated by PIV technique combined with microscope. The relationship between the flow characteristics upon closure and the transient pressure drop was thus studied and discussed.

6.1 Transient Pressure Measurement

The pressure field upon occluder closure is quite complex, involving fluid deceleration, squeeze flow, solid impact between the occluder and housing, acoustics and possible cavitation occurrence. An important phenomenon at the MHV occluder/housing impact is the negative pressure transient (NPT) occurrence at the inflow side. The NPT is induced by two possible mechanisms: 1) a tension wave of reduced pressure created by the sudden stoppage of the flowing fluid upon the occluder closure, commonly known as the water hammer effect; 2) a vortex created with low pressure core when the high speed squeeze jet rushing out the gap between the approaching occluder and the housing, and mixing with the ambient fluid.

The NPT on the inflow side upon MHV closure is an important aspect in cavitation study. It has been intensively investigated in a pulsatile mock circulatory flow loop (Garrison et al., 1994; Wu et al., 1995a; Sneckenberger et al., 1996), single closing burst tester system (Lee et al., 1996a; Lee et al., 1996b; Chandran et al., 1997; Chandran et al., 1998a) and in vivo (Chandran et al., 1998b; Dexter et al., 1999; Paulsen et al., 1999). All of these measurements demonstrated an abrupt fall of pressure below the liquid vapor pressure at the instant of closure, suggesting a cavitation potential both in vitro and in vivo. The NPT was followed by high frequency pressure oscillation (HFO), which can be attributed to the collapse of cavitation bubbles (Hwang 1998; Chandran et al., 1998b). The analysis of the HFO's power spectrum showed its frequency were in the range of 50-200k Hz (Chandran et al., 1998b; Lin et al., 2000).

Comparing to the mitral valve, the aortic valve closure motion in a physiological mock flow loop is driven by less pressure gradient and thus lower occluder impact velocity is expected. It is therefore suggested that the aortic valve has less opportunity to cavitate under the same conditions. However, for the current test loop setup, squeezing flow measurement is difficult to carry out around a mitral valve by PIV technique. Thus the aortic valve instead is selected to study the relationship between squeeze flow and cavitation. The investigation of the aortic valve's closing dynamics is believed to provide some insights to the understanding of the cavitation mechanisms in MHV.

In this work, NPT with different MHV models (tip geometry) was recorded, together with the closure flow field measurement. The cavitation implication for different valve tip designs can thus be compared and correlated to the MHV flow dynamics upon fully closure. The outflow side pressure was also simultaneously investigated.

6.1.1 Different Valve-tip Geometries and Locations of Pressure Transducers

Fig.6.1 shows three MHV models with different valve-tip geometries at their fully closed positions. These valve models feature the tip angles of 30, 60 and 90 degree, but their dimensions and opening & closing angles are the same. The valve-tip angle of 60° is supposed to close with all the valve full contact with the housing wall. However, for 30° and 90° geometries, only tip contact is possible. Although the two designs of 30° and 90° are not practical, this study was introduced to understand the effects of such configurations. The theoretical dimensions of OA and OB (point O being the intersection of the valve ring with the inner pipe housing), as in Fig.6.1, were calculated as 1.975mm and 3.13mm respectively. Two identical pressure tapping points, H1 and H2 were drilled through the aortic pipe wall, inside which two Millar micro-tip pressure catheters with diameter of 1mm were installed. The diameter of the pressure tapping points were 0.5mm larger than that of the pressure catheter, so that the fluid will be in full contact with the sensing element. The catheter tip should not extend into the flow field to avoid the unexpected disturbance for the squeezing flow measurement. The distance from the catheter center to the two end tips upon fully closure, point A and B is around 1.3mm. Theoretically, the thinner of the catheter tip diameter, the more accurate of the pressure measurement for the specific point. Also its existence causes less interference for the original flow field. However, pressure signal is normally poor for most micro sensing elements less than 1 mm under current technology. Moreover, the orientation between the transducer and flow direction may also influence the measurement results. The optimization of the tip diameter selection and its position & orientation need to be further studied.

(a) fully closed occluder with 60° valve tip

(b) fully closed occluder with 30° valve tip

(c) fully closed occluder with 90° valve tip

Fig. 6.1 Schematic sketch for different valve tip designs upon fully closure

The transient pressure change on both the inflow and outflow side of the MHV were measured using two micro-tip pressure catheters (SPC-330A, Millar Instrument Inc., USA). The SPC-330A catheter has a tip diameter of 1mm and a natural frequency over 30 KHz and its overpressure ranges from -760 to 4000 mmHg. The transducer was calibrated prior to use, according to the Millar instrument manual (Micro-tip catheter pressure transducer human use: instruction for use) and its specifications are listed in Appendix III.

The catheter pressure signal was first amplified by a transducer control unit (Model PCB 500, Millar Instrument Inc., USA) and then measured and recorded by an AR1200 Analyzing Recorder (Yokogawa, Japan). The analyzing recorder is equipped with 8 isolated analog input channels and 14 bit A/D converter (maximum sampling rate of 100kS/s). Its memory capacity was 64K words /channel. The pressure signal was sampled

with sampling rate of 20 KHz using this instrument and then transferred to a desktop computer for further processing. In this part of experiment, the volumetric flow rate was raised to 4.3 liter/min. Fifty cycles of the two channels pressure upon MHV closure were simultaneously measured, the mean NPT and peak outflow values were calculated respectively.

6.1.2 Transient Pressure Results

The NPT recording was triggered immediately before valve fully closed. The triggering instant for the pressure measurements for the three different valve tip designs may differ due to the photoelectric sensor relocating in the experiment process when changing the valve models. Thus the starting point may differ for measurements of occluders' pressure wave traces. Typical pressure waveforms for the tip angles of 30°, 60° and 90° are shown as Fig.6.3 (a), (b), (c). These results demonstrate that both the inflow and outflow pressure experience dramatic changes upon MHV closure with duration around 1ms. There are normally two largest pressure drops on the inflow side, while for the outflow side, positive transient pressure change is observed. For comparison purpose, two parameters as $P_{\max 1.inflow}$ and $P_{\max 2.inflow}$ from each inflow wave trace, one parameter $P_{\max.outflow}$ from outflow pressure waveform are identified, as illustrated in Fig.6.2. The mean values of $P_{\max 1.inflow}$, $P_{\max 2.inflow}$ and $P_{\max.outflow}$ over 50 cycles for different tip angles are summarized in table 6.1.

Table 6.1 Average pressure rise and drop at different tip angles

Valve tip angles	Inflow side		Outflow side
	$P_{\max 1.\text{inflow}}(\text{mmHg})$	$P_{\max 2.\text{inflow}}(\text{mmHg})$	$P_{\max.\text{outflow}}(\text{mmHg})$
30 degree	-253	-320	94
60 degree	-312	-183	91
90 degree	-283	-151	104

Fig.6.2 Illustrative definition of the transient pressure drop and rise

(a) 30° valve tip

(b) 60° valve tip

(c) 90° valve tip

Fig.6.3 Transient pressure at inflow and outflow upon closure for different valve tip designs

Among the three different valve tip, the occluder with 30 degree tip angle demonstrates the largest magnitude for the second pressure drops and minimum magnitude of the first NPT

on the inflow side. On the other hand, the 60° tip design is found with the maximum value for the first transient pressure change. It is also shown that for 30° tip design, the second NPT is larger than its first one, but for the other two designs, the opposite trend is observed. The second peak pressure drop for the 60° and 90° tip designs is dramatically reduced by 40-55% comparing to that in 30° valve tip. The outflow pressure rise for the 30° and 60° valve tip shows the similar magnitude, and the value for MHV with 90° tip angle is marginal.

6.1.3 Discussions

The NPT peaks shown on the inflow side of all the three valve models is due to consecutive phenomena associated with occluder closing dynamics, including fluid acceleration in the pre-squeeze flow phase, squeeze flow, water hammer effect and/or occluder rebound, etc. From the temporal domain, the first NPT peak is likely to be related with the first effect, i.e., pre-squeeze and squeeze flow phenomena, while the second NPT peak may be associated with the second effect, i.e., solid impact, water hammer effect and possible rebound motion. For 30° tip design, the NPT induced by the second effect is stronger than by the first effect. Whereas for 60° and 90° tip occluders, the first effect is more important for NPT formation. The three different valve designs shown the maximum NPT value around -300 mmHg on the inflow side. This condition will not initiate cavitation since the pressure is not sufficiently low as fluid vapor pressure (-740 mmHg at 25° C for water). However, it can be expected that lower pressure can be obtained in the mitral position and the valve occluder is PyC material. And in some extreme cases, the pressure may possibly be low enough to induce cavitation.

For the transient pressure rise induced by fluid compressibility on the outflow side, various valve tips shows negligible difference. The notable pressure increase on the outflow side vs the sharp pressure drop on the inflow side can cause high leakage flow after the occluder is fully closed (Maymir et al., 1997; Steegers et al., 1999; Aluri et al., 2001). Large wall shear stress may thus occur and cause damage to the blood cells and/or activate the platelet that results in hemolysis & thromboembolism.

6.2 Closure Flow Measurement

PIV experiment was repeated for the three different valve-tip designs. The magnification for this series of PIV experiments was chosen as 3.2 and the size of the field of view (FOV) was around 2.7 mm. Due to the different geometry of the occluder tip, X and Y coordinate in the image field may not be identical if the fully closed occluder was finely tuned to locate in the middle of the PIV picture.

For the purpose of description, various edges of the occluder are designated as in Fig.6.4.

The gap size is the distance between the housing wall and the closer wall occluder tip.

Fig.6.4. Designations of occluder edges

6.2.1 Gap Around 0.5 mm

Fig.6.5 shows the flow field and vorticity maps for the three different valve-tip designs when the gap is around 0.5mm. Some similar flow features are demonstrated in the upstream and downstream region for all the types of occluders. The two main features include:

- 1) the occluder is pushed to close by the backflow in the downstream with the flow velocity perpendicular to the leaflet surface;
- 2) the flow in the upstream is not simply induced by the occluder rotation, as its direction is not perpendicular to the occluder surface, but parallel to the axial direction. It is therefore suggested that the upstream flow is caused by the combined effects of the momentum transfer from the occluder's displacement, squeeze flow from the gap and the occluder translation movement. The occluder translation is due to the clearance in the hinge, which is necessary for the free rotation of the occluder.

The three types of occluders are also found with similar velocity magnitude both in the upstream and downstream region, with the maximum velocity of 0.76~0.81 m/s. The major flow difference for the various tip geometries is the point where the flow divides into two streams on the topside. One stream is pushed by the occluder movement and the other one is the wake induced by the drag force. For the 30° valve-tip, as in Fig.6.5(a), the two stream separates close to end of the topside, while this separation locates along the midsection of the topside for 60° valve tip, in Fig.6.5(b). The 90° tip valve tip, as in Fig.6.5(c), on the other hand, has a vortex formed on the topside.

From the vorticity map, it is shown the relative large vorticity is found around the topside, occluder boundaries. The vortex strength at the sharp end of 30° tip occluder is stronger than the other two geometries.

6.2.2 Gap Around 0.35 mm

With the gap decreases to 0.35mm as shown in Fig.6.6, stronger backflow in the downstream region pushes the occluder to close and the occluder accordingly gives more momentum to the upstream flow. The main flow features and topside flow characteristics mentioned in the gap 0.5mm still maintains. The velocity magnitude increases both in the upstream and downstream regions. Vorticity strength in the field also increases, suggesting the rise of velocity gradient with the gap decreasing. The 30° tip occluder has the largest vortex intensity.

(a) 30° valve tip

(b) 60° valve tip

(c) 90° valve tip

Fig.6.5. Velocity and vorticity maps when gap is 0.5 mm

(a) 30° valve tip

(b) 60° valve tip

(c) 90° valve tip

Fig.6.6 Velocity and vorticity maps when gap is 0.35 mm

6.2.3 Gap Less Than 0.1mm

When the gap continuously decreases to less than 0.1 mm, the flow maps for the three different valve-tip designs are shown in Fig.6.7. For the case of 60° valve tip, as in Fig.6.7(a), high speed jet is found squeezing outside the gap and mixes with the relative low speed fluid at the inflow region. Whereas for the 30° and 90° valve tip, it is less possible to observe the real squeeze flow phenomenon. Although their maps shown in Fig.6.7(a) and (c) have than gap dimension less than 0.1mm, the flow speed at the exit of gap does not demonstrates much higher magnitude comparing to the nearby region. Their flow structures actually inherit the characteristics possessed by those with larger gap dimensions. The maximum flow velocity is found occurring at the leading tip of the occluder, with the value around 1.3~1.4 m/s.

For the case of 60° valve tip, the squeezing jet from the gap is found around 2.1 m/s, larger than the magnitude for 30° and 90° valve tip. After the squeezing jet rushes out of the gap, it mingles with the low momentum flow region. Stronger velocity gradient and vorticity are therefore formed in this inflow region. The 60° valve tip is also found with larger vorticity value at the inflow side comparing to the 30° and 90° tip occluder, suggesting a lower pressure is formed in its vortex center.

(a) 30° valve tip

(b) 60° valve tip

(c) 90° valve tip

Fig.6.7 Velocity and vorticity maps when gap less than 0.1 mm

6.2.4 Occluder Sudden Stop

Immediately after squeeze flow, the occluder comes to a sudden stop as shown in Fig.6.8. The occluder is still pushed by the backflow, but the driving action is dramatically reduced, with velocity around 0~0.15 m/s. The main backflow shifts to the occluder tip and a vortex is formed there. The flow at the inflow side is also sharply decreased its magnitude due to the greatly reduced momentum transferred from the occluder upon its sudden stop. The 90° occluder features relative higher flow velocity of 0.4~0.5 m/s at the inflow, in Fig.6.8(b), comparing to that in the 30° occluder with value around 0.2 m/s as shown in Fig.6.8(a). The water hammer pressure induced by the occluder sudden stop is produced at the upstream side, which may facilitate the cavitation occurrence.

(a) 30° valve tip

(b) 90° valve tip

Fig.6.8 Velocity and vorticity maps upon occluder sudden stop

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6.2.5 Occluder Rebound

Immediately after the occluder sudden's stop, there is sometimes a rebounding phenomenon. Fig.6.9 shows typical rebound flow maps for different valve tip designs. Comparing to the upstream flow for the scenarios of squeeze flow and occluder sudden stop, the flow at the inflow side now starts to reverse its direction, which impacts on leaflet surface forcing it rebound. The 30° occluder demonstrates relatively larger reversed flow with the magnitude around 0.4 m/s, one time larger than those for the other two types of occluders. A vortex is also found around the occluder upside resulting from the combination of reversed flow and still existing weak forward flow in this region.

The flow adjacent to the occluder downside is no long perpendicular to the surface upon rebounding. Instead it is now a mixture of parallel and vertical flow near the leaflet surface, resulting from the combined effect of occluder rebound motion and the backflow from the aorta. The flow in other downstream regions is mainly influenced by the backflow.

(a) 30° valve tip

(b) 60° valve tip

(c) 90° valve tip

Fig.6.9 Velocity and vorticity maps upon occluder rebound

6.2.6 Discussions

The flow maps under different phases, including pre-squeeze flow, squeeze flow, occluder sudden stop and rebound are presented for the three different valve-tip designs under increased flow rate of 4.3 liter/min. The occluder rebound action is more likely to occur with increased flow rate, as well as the more dramatic phenomenon of NPT pressure traces. The consecutive events of squeeze flow, occluder sudden stop and rebound are transient phenomena, which occur in the order of microseconds. Thus large numbers of PIV pictures are required to capture and isolate these closely successive motions especially for larger flow rate, since the time duration for completing all these events is small. A triggering mechanism is therefore necessary to start the PIV system, when the occluder is near to the closure position. A time step delay of $10\ \mu s$ was introduced for fine-tuning.

It is known that the occluder closure phase is an unsteady, turbulent phenomenon, and the turbulence intensity is especially large when the occluder is near fully closing under the complicated reactions of squeeze flow, sudden stop and rebound. The maximum Reynolds number for this localized, millimeter dimension flow field is calculated as 590, if the occluder thickness, 1mm is selected as the characteristics length.

Before squeeze flow occurring, the three different valve tip designs show similar flow magnitude and characteristics in the upstream and downstream regions. The overall flow velocity increases in comparison to those in Chapter 5 with flow rate of 3.2 liter/m. The backflow keeps perpendicular to the occluder backwind surface and drives the occluder to close. The upstream flow keeps parallel to the axis direction and this parallel flow is believed to be the resultant effect of gap flow, occluder rotation and translation.

The occluder squeeze flow, sudden stop are differentiated by different flow features. The squeeze flow is characterized by the high speed jet ejecting from the gap between the occluder and the housing wall. The squeeze flow is closely followed by the occluder's sudden stop. When the occluder comes to a stop by impacting on the housing wall, the flow adjacent to the occluder downside surface is sharply reduced near zero with its direction still perpendicular to the surface. The flow velocity also dramatically drops at the inflow side upon occluder sudden impacting. Immediately after its stop, the occluder rebound occurs. Reverse flow starts to emerge at the inflow side to push the occluder rebound. Relative disorderly flow is found close to the occluder downside surface due to the rebound motion and the effect of the nearby backflow.

For the abovementioned flow phases, different valve-tip designs demonstrate their tip design influence for the flow maps. For the pre-squeeze flow phase, one notable different flow feature among the three valve-tip geometries is the variable positions where the wake is formed around the leaflet. The sharper the occluder tip, the later the vortex emerges along the leaflet topside. The vortex intensity also decreases as the tip angle becomes larger.

Comparing with the 60° valve tip design, the 30° and 90° valve tip have a less tendency for the “squeeze flow” occurrence. Due to their sharp tip geometries, the gap structures play little role in speeding up the backflow inside the gap, which may lead to the relative low magnitude of the flow velocity at the leading tip of the occluder. The 60° valve tip, on the other hand, is observed with large squeezing velocity at the immediate exit of the gap. The strong vorticity is also observed for this occluder, implicating that it is subjected to more pressure drop at the inflow region.

For the occluder sudden stop, the 90° occluder features relative higher flow velocity, 0.4~0.5 m/s, at the occluder tip in the inflow comparing to that in the 30° occluder, whose value is around 0.2 m/s. One possible explanation is that the gap structure of the 90° occluder still enables the existing of the weak backflow to the inflow side through the gap, which is contrary to the scenario of the 30° valve tip. The dramatic velocity decreased from “squeeze flow” to sudden stop is completed within order of microseconds. Large pressure drop is thus initiated due to this high flow deceleration rate at the inflow side.

For the occluders rebound, the flow at the inflow region starts to reverse and push the occluder to open. The flow velocity in this area is relatively lower for the 60° and 90° occluders due to the resisting backflow inside the gap. The occluder rebound further expands the magnitude of the flow deceleration rate and increases the pressure drop inside this region. The 30° occluder is thus considered to have the largest pressure drop from the water hammer phenomenon and occluder rebound.

6.3 Conclusions

MHV closure phase is an unsteady, turbulent flow field, which is difficult to simulate accurately through the CFD study. The phenomenon associated with the occluder final closure dynamics, including squeezing flow, sudden stop and possible rebound are especially complicated for CFD computation due to the fluid-structure interaction inside the minute dimension, inhomogeneity and unsteadiness, etc. Experimental work is therefore needed to provide some insights for this complicated phenomenon and also verifications for the future CFD results.

Three different valve tip designs were studied in this work. NPT is found for all the valve models and different pressure drop magnitude is demonstrated for the various models. It is shown that the 30° occluder has the largest magnitude of the second NPT peak at the inflow side, while the 60° is observed with maximum value of the first NPT peak.

The occluder near-closure flow phenomenon investigated by PIV technique and the results can be subdivided into pre-squeeze flow, squeeze flow, occluder sudden stop and rebound in terms of their respective characteristics. Each flow phase is influenced by different valve tip designs. The pressure drop associated with squeeze flow, valve stop thus also differs for various valve models. The PIV results demonstrate that the 60° occluder has higher squeezing flow velocity, which may results in its largest magnitude of the first NPT value. On the other hand, the 30° occluder is observed with more evident velocity change at the occluder sudden stop and rebound, which may also be the cause for its second NPT being the maximum among the three designs.

This experimental study shows that the different valve tip designs influence the occluder closing phenomenon, including squeeze flow, occluder sudden stop and rebound. NPT occurring at MHV inflow side is found related with the closing dynamics and its magnitude is also influenced by the valve tip design.

Chapter 7 Conclusions and Future Works

7.1 Conclusions

Knowledge of flow fluid in MHVs is of interest due to the link between the fluid dynamics characteristics and complications associated with the valves' performance. It has also been believed that the damage of blood depends on the local shear stress, while thromboembolism is promoted by the interplay of the stressed blood cells and flow stasis or recirculation around the MHV region. Although much work has been carried out to study the pulsatile flow past MHV using various techniques, detailed investigation coupled occluder dynamics & flow features, and stress distribution across MHV structure is limited. In vivo cavitation in MHV has received interest and attention among researchers, manufacturers, and regulatory agencies since the discovery of cavitation-like erosion and pitting in pyrolytic carbon components in a few of ED MHV clinical explants. Several basic fluid mechanical factors have been implicated to induce the occurrence of MHV cavitation, but the exact mechanism has not been well understood. Based on the fundamental process of cavitation occurrence and previous cavitation studies, it was hypothesized that the squeeze flow created during the final phase of a valve closing, which is affected by the detailed geometry of approaching boundaries, plays an important role in the potential of MHV cavitation.

In this study, a pulsatile flow test rig was specifically designed and fabricated to investigate the global and squeeze flow phenomenon in a physiological circulatory flow system. An actual size complete transparent bileaflet MHV model was developed to facilitate the flow field measurement through the lumen, as well as through the gap between the occluder and the housing wall using PIV technique. A precise phase lock method, which acquired the

PIV trigger signal from the occluder motion, was developed to resolve the temporal resolution of the pulsatile flow in one cardiac cycle, and particularly the squeeze flow within the final microseconds of the valve closure by a time step of $10\mu\text{s}$. A water, glycerin and sodium iodide solution was used as the blood analog, which had the similar viscosity of blood at relatively high shear rates, as well as the matching refractive index of test section to minimize the optical deformation in PIV measurement. The global flow through the occluders was analyzed in one complete cardiac cycle, together with the measurement of the leaflet dynamics and shear stress distribution. For the squeeze flow study, a microscope was incorporated with the PIV system to enlarge the narrowing channel flow with dimension around 1.4 mm to less than 0.1 mm.

Additional two sets of transparent bileaflet MHV models were fabricated to investigate the influence of different valve tip designs for the transient pressure and closure flow features. The transient pressures upon closure were recorded using micro-tip pressure catheter in the vicinity of the various valves and their squeeze flow fields were also compared.

The following conclusions can be drawn from the results of this study:

- 1) The global flow field across MHV was separated into three channels by the presence of two moving occluders, namely one central channel and two side channels. The flow momentums inside three channels kept increasing till the valve fully opened. After peak systole, the forward flow velocity started to decrease until back flow appeared, which pushed occluder to close. At every specific instant of the opening process, the occluder position demonstrated good repetitiveness between various cycles. The occluder opening motion was characterized with decreasing velocity, due to the stronger resistance action of the central jet expansion. The shear stress was found increasing in the opening process and

decreasing in the fully opened stage. The peak value reached 500 N/m^2 at peak systole, which is larger than the threshold for hemolysis occurrence. The maximum stress was normally observed around the hinge area inside the central channel, where the central jet separation caused the strongest turbulent mixing. The wake of the occluder tip was also demonstrated to have large shear stress, which was in agreement with other investigations of the MHV downstream flow field measurement.

2) MHV closing process was initiated by the back flow, which was strengthening with the acceleration of the occluder closing process. The closing shear stresses were found less than those in the opening process, due to the relatively weak regurgitant flow.

3) For the squeeze flow study, series of images within the continuously narrowing flow channel were captured. As the gap between the tip of the valve occluder and orifice wall becomes narrower, evidence of high speed jet flow becomes more apparent. When the flow channel was reduced to around 0.1 mm , flow velocity of up to 2 m/s is noted, with the strongest vorticity existing near the leading edge tip of the occluder. This sudden increase in high-speed jet flow would be accompanied by a corresponding reduction in local pressure, and hence a likely source for potential cavitation.

4) Negative pressure transients (NPT) with different magnitudes of two peaks were demonstrated for three different valve-tip designs upon closure in the inflow side. Incorporated with the closure field measurements, the valve tip 60° was shown with the highest squeezing flow velocity, which may results in its largest first NPT peak. The 30° valve tip, on the other hand, was observed with the most evident velocity change upon the occluder sudden stoppage and rebound, which was considered to produce the maximum second NPT peak value among the three designs.

7.2 Future Works

The current study mainly focus on the in-vitro measurement of global and squeeze flow across a MHV model. The influence of the valve tip design for the squeeze flow pattern and transient pressure reductions was also investigated. Future works are proposed as follows:

- 1) To design an efficient experimental technique for the observation of the cavitation bubble formation and collapse in the vicinity of the MHV, along with the measurement of the flow dynamics upon valve closure.
- 2) By neglecting certain factors that secondary to the cavitation potential, computational fluid dynamics (CFD) approach can be applied to model the local transient flows, which can be compared with the experimental results.
- 3) The effect of MHV design factors need to be further studied with more realistic modifications on occluder/housing contact geometry, thus to provide more complete and practical suggestions for the future valve design.

Summary of Achievements

Cavitation in MHVs has been well investigated since 1994. Several experimental computational fluid dynamics investigations were carried out. However, the flow physics in relation to cavitation has not been studied in such details as in this research project.

The main achievements in this PhD research can be summarized as follows:

- A) A novel triggering technique was designed and applied in this investigation. This allows the motion of the valve occluder to be tracked and controlled the particle image velocimetry (PIV) operation with a time delayed program. Thus the dynamics of the transparent valve occluder and nearby flow field can be simultaneously investigated. More importantly series of squeezing flow field immediately before MHV closure was captured in details. Such flow details was not investigated previously.

- B) The stress distribution within the lumen of the two occluders were measured at every specific instant of the cardiac cycle, including the opening process, full opened stage and closing process. In particular, the hinge area, where the central jet separates from the occluder surface was found with the largest shear stress, due to the strong turbulent mixing.

- C) A microscope incorporated PIV system was used to study the continuously narrowing channel flows before the occluder fully stoppage. For the first time, the evolvments of the velocity distribution in the occluder final closing phase, from the dimension around 1.4 mm to less than 0.1 mm, were experimentally demonstrated. The

squeeze flow was finally observed when the gap is less than 0.1 mm, which may produce favorable condition for the cavitation occurrence.

D) Three sets of MHV models with different occluder tip designs were applied to study the valve tip influence for the closure flow field and negative pressure transients (NPT), thus establishing a link between the MHV design, squeeze flow and cavitation potential. The closure flow field, including squeeze flow, occluder sudden stop and rebound was measured for each MHV model and correlated with NPT recording.

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Appendix I Operational Settings of Photoelectric Sensor

● Light axis adjustment

Light axis adjustment can be selected in SET mode.

The strength of light which enters into the amplifier unit is indicated from 0 to 100.

Adjust the fiber light axis to get the maximum level.

In this case, no output is generated.

● Sensitivity Setting

Appendix II

Properties of model material and refractive index matching fluids in PIV experiment

Properties of model materials

Material		Model fabrication methods	refraction index	Scratch resistance	comments
Glass	Pyrex glass	Glassblowing , fusing	1.47~1.49	+++	
	Optical glass	Grinding , polishing	1.45~1.96	+~++++	
Plastic	Acrylic perspex	Machining, bonding, bending, polishing	1.49	+	
	Polycarbonate stinging resin	Machined after casting except silicon	1.58	++	
	Epoxy based		1.56	+	
	Ethane based		1.49	++	Cure at 65°C
	Acrylic based		1.49~1.53	+	
	Elastimer		1.43	+	Elastic walls

Excerpted from Budwig, R (1994).

Properties of refractive index matching fluids

Matching fluids	n Refractive index	$\frac{\rho}{\rho_0}$	$\frac{\mu}{\mu_0}$	References and comments
Aqueous solution				
Glycerin	1.33-1.47	1-1.26	1-1490	
Zinc iodide	1.33-1.62			Hendricks & Avram(1981)
Sodium iodide	1.5, 60% NaI by wt.			Chen and Fan (1992)
Potassium thiocyanate	1.33-1.49	1-1.39	1-2.4	Jan et al.(1989), with glycerin to increase viscosity
Ammonium thiocyanate	1.33-1.50	1-1.15	1-2.1	Hopper (1992)
Sodium thiocyanate	1.33-1.48	1-1.34	1-7.5	Duncan et al.(1990)
Organic liquids				
Kerosene	1.45	0.82		Bovendeered et al.(1997)
Silicon oil mixture	1.47	1.03	190	Edwards & Dybbs(1984)
Mineral oil	1.48	0.85		
Turpentine	1.47	0.87	1.49	
Solvent naphtha	1.50	0.67		
Soybean oil	1.47	0.93	69	
Olive oil	1.47	0.92	84	
Castor oil	1.48	0.96	986	
Tung oil	1.52	0.93		
Cassia oil	1.60			Sinkankas(1966)
Dibutylphthalate	1.49			

Excerpted from Budwig, R. (1994). ρ_0 and μ_0 are the density and absolute viscosity of water at 20°C.

Appendix III

3F - 10F CATHETERS MIKRO-TIP TRANSDUCER SPECIFICATIONS STANDARDIZED SENSITIVITY

Type of Sensor	Diffused semiconductor
Pressure Range	-50 to +300 mmHg (-6.7 to 40 kPa)
Overpressure	+4000 mmHg (+530 kPa) -760 mmHg (100 kPa)
Rated Excitation	2.5 - 7.5 V _{dc} or V _{ac} , rms*
Sensitivity	5 μ V/V/mmHg, nominal (37.6 μ V/V/kPa)
Temperature Error Band at Zero Pressure	$\pm$ 1 mmHg ($\pm$ 0.13 kPa) BSL, 23 - 38 ° C
Linearity and Hysteresis (Combined)	$\pm$ 0.5%, BSL of full scale
Drift **	Less than 6 mmHg (0.8 kPa) in 12 hours
Natural Frequency	$\geq$ 35 kHz
Bridge Resistance	1000 ohms, nominal
Reference Pressure	Atmosphere
Electrical Leakage	Less than 10 μ A at 500 V _{dc}
Zero Offset	Less than $\pm$ 50 mmHg ($\pm$ 6.7 kPa)

* Performance specifications are for 5 V_{dc}. Transient voltages up to 20 volts will not damage the transducer.

** After zeroing in water. For maximum stability, presoak transducer in sterile water or saline for 30 minutes prior to use.

($\pm$ 0.1 kPa)

* Performance specifications are for 5 V_{dc}. Transient voltages up to 20 volts will not damage the transducer.

** After zeroing in water. For maximum stability, presoak transducer in sterile water or saline for 30 minutes prior to use.