Transient contour of bioprosthetic heart valve leaflet : an experimental and numerical investigation

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Transient Contour of Bioprosthetic Heart Valve Leaflet: an Experimental and Numerical Investigation

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Abstract

Bioprosthetic heart valve (BHV) is known to have a limited life span of between 6 to 14 years. The leading cause of its failure is the stresses in the valve leaflet. In order to improve the durability of the BHV, the treatment of the valve and its design are critical. Understanding the transient contour of the stress in the BHV is essential to realize the optimal design of BHV.

Numerical simulation has been an effective approach to achieve this. However, the current finite element method (FEM) for the analysis of the BHV has several shortcomings; it is deficient in term of the capability in modeling the realistic boundary conditions such as the physiological pressure loading, the material properties, and the geometrical profile of the moving valve leaflet. Moreover, the existing FEM could not describe the natural opening/closing process of the valve leaflet and its movement in the fluid. This PhD project aims to address these shortcomings by developing an experimental technique to acquire three dimensional images of an aortic valve leaflet during the opening and closing process. In addition, a novel dynamic FEM model of the BHV would be built using the data from this experimental technique.

Hence, the main contribution of this PhD research is a novel engineering technique to obtain a more realistic boundary conditions as input, and hence a more accurate simulation model, results, and interpretation.

The experimental technique was developed using an optical and image processing techniques to track the transient contour of the moving valve leaflet. A dual-camera stereo photogrammetry (DSP) technique is proposed and it comprises of the two phases. In the first phase, two identical cameras are focused on the BHV leaflet to acquire the leaflet images simultaneously from two different angles. In the second phase, the DSP is used to retrieve the 3-D coordinates of the marker matrix drawn on the leaflet surface

via pairs of instantaneous digital images during a complete valve opening and closing cycle. A structure with known dimension was used to calibrate the camera set up.

After the experiments have been conducted, the dynamic leaflet motion was reconstructed from a sequence of images. The images of the same marker were identified from its relative location in the marker matrix pattern, and these images were then spatially paired for the 3-D coordinate calculation.

When the 3-D locations during the cardiac cycle had been reconstructed, a numerical model was created. The BHV in a relaxed state was chosen as the initial geometry of the numerical model. Besides the pressure load applied on the leaflet surface, the 3-D displacements of the free edge (retrieved from the experiment) were also applied as dynamic boundary conditions in the numerical model.

There has been no realistic modeling method for the BHV that combines the numerical simulation with the experimental data. In this PhD work a new technique is proposed, in which the numerical simulation was correlated with the experiment work; simulating more accurate physiological deformations using experimental measured data. This new modeling technique can show the sway of the leaflet during the cardiac cycle, which is comparable with the sway observed in the real valve.

The main objective of this PhD thesis has been achieved. A novel engineering technique involving capturing of dynamic 3-dimenisonal data of the valve leaflet in opening and closing has been successful carried out. These data were input as boundary conditions to a FEA software and illustrated more realistic results. This technique is particularly useful the next generation heart valve design, namely the stentless valve.

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Chapter 1 Introduction

1.1 Background

Human heart can be considered as a two-stage pump. The right side of the heart receives blood from the veins and pumps it through the lungs for oxygen exchange, while the more powerful left side ejects the oxygenated blood through the aorta to the body tissue and organ. The right side of the heart comprises of two chambers namely right atrium and right ventricle. And similarly, left atrium and ventricle for the left side of the heart. Blood flows through these chamber via a series of valves. The function of a heart valve is to control the direction of flow, in and out of the chamber in one direction only. The valve prevents the backward flow of blood. There are four valves in the heart, namely :

tricuspid valve	located between the right atrium and the right ventricle
pulmonary valve	located between the right ventricle and the pulmonary artery
mitral valve	located between the left atrium and left ventricle
aortic valve	located between the left ventricle and the aorta

If a heart valve is congenitally defective or experience deterioration, then blood flow across the valve can be adversely affected and leading to reduced cardiac output. In order to maintain the required cardiac output, the heart muscles response by pumping harder to overcome the reduced cardiac output. This causes ventricle enlargement and eventually leads to heart failure.

Since the aortic valve undergoes highest loading, it is often affected by diseases, resulting in severe consequences (Schoen FJ, 1999). The diseased valve may be repaired or replaced with a valve substitute. Although valve repair is generally superior to valve replacement, only about 5% of these can be repaired. If the valve cannot be surgically repaired, the only alternative is the replacement of the defective valve with an artificial heart valve.

In general, prosthetic valves can be divided into two groups: mechanical heart valves (MHVs) and bioprosthetic heart valves (BHVs). MHVs are made with 100% synthetic materials and are known for their durability, but the patient would have to take anticoagulation drug for life, to prevent thrombus formation. Mis-management of anticoagulation drugs can lead to severe complications such as hemorrhage.

BHVs on the other hand, are made with chemically treated animal tissue; such as valve from pig or using pericardium from young calf and reconstructed into a valve. Compared to MHVs, the BHVs behave closer to the natural heart valves. However, these BHVs degenerate and has a limited life-span ranging from 8-15 years. Calcification and leaflet wear are the most common associated degenerations that lead to BHV failure and when this happen, re-operation is needed. An ideal artificial valve is not available at the moment. The next generation replacement valve will most likely be a tissue engineered valve, using living cells.

Tissue engineering applies engineering principles to biological systems to create viable structures for replacing diseased or deficient natural organs (Breuer et al. 1996). It has large potentials in developing a more durable artificial heart valve with similar mechanical behavior as the BHV. Therefore, a thorough investigation of the mechanical behavior of the BHV, especially its failure mechanisms, will not only improve the current BHV designs, but also serve to guide the development of the tissue engineered heart valve in the future.

Many evidences indicated that the failure of the BHV leaflet was most likely related with the mechanical stresses in its material (Gabbay S et al., 1986,1988; Vesely I et al., 1988, 1989). Though many have studied the relationship between the leaflet failure and the mechanical stress in BHVs, apparently none investigated the true leaflet motion (large material deformation). One of the difficulties in numerical studies is to develop an appropriate material model to describe the complex property of leaflet tissue. There is also no numerical research done that consider the dynamic deformations of the leaflet.

On the other hand, the experimental method suffers from the lack of the universal technique that provides a complete solution to the problems associated with the valve motion. For instance, most of testing techniques can only account for the surface deformation of the leaflet. In addition, it is hard to perform an experiment that can mimic the in-vivo state of the heart valve. Therefore, a hybrid experimental-numerical approach is adopted. In this approach, the numerical analysis is correlated with experimentally determined parameters, and the result is validated by experiment. On the other hand, the experiment is also designed to be consistent with the numerical model.

1.2 Objective

The main objective of this PhD project is to develop a numerical dynamic simulation during the entire cycle using an experimental dynamic leaflet profile as additional boundary conditions. This will provide a more realistic mechanical simulation of the aortic valve. There are several crucial factors for a successful numerical simulation; they are finite-element types, material properties, loadings and boundary conditions. In the state-of-art biomechanics numerical analysis the boundary conditions are still not adequate to model the real valve. Therefore, the current PhD research aims to improve the boundary conditions of the BHVs (by obtaining them from the experiment) to compensate the lack of information on the fluid loading. Hence, a more physiologically-real deformations and stress distributions can be achieved.

Thus, the main contribution of this PhD research is a novel technique to obtain a more realistic boundary conditions, and hence a more accurate simulation model, results, and interpretation.

1.3 Scope

The scope of this research is divided into the following phases:

- a) Development of an experimental setup capable of capturing physiological leaflet motion as it opens and closes under pulsatile physiological pressure
- b) Reconstruction of the 3-D coordinates of the moving leaflet from the paired 2-D images captured during the experiments
- c) Development of a numerical model of the tested BHV leaflet
- d) Correlation between the numerical model and the experimentally measured 3-D leaflet dynamic-profiles
- e) FE model simulation with the dynamic 3-D leaflet profiles as boundary condition to investigate the mechanical behavior of the valve leaflet
- f) Comparison with the existing BHV leaflet model

Chapter 2 Literature Review

2.1 General Introduction to Heart and Heart Valve

2.1.1 Anatomy of the Human Heart and Its Function

The human heart (Figure 2.1) is the core of the cardiovascular system. Its main role is to pump blood. The left side takes oxygenated blood from the lungs and pumps it out to the whole body through the aorta. After exchanging oxygen with body cells, the blood running back to the right part of the heart through vena cava (systemic circulation). Next, the heart will pump the deoxygenated blood through the pulmonary artery to the lungs to be oxygenated (pulmonary circulation). This cycle then repeats.

Figure 2.1 The human heart and its valves (www.cts.usc.edu)

The heart is divided into two sides (the left and the right) to keep the two types of blood (oxygenated and deoxygenated) from mixing. Each side of the heart has an upper half (the atrium) which works as a “priming pump” to fill the much stronger lower half (the ventricle) which does the stronger pumping action. The right ventricle, which is relatively smaller than the left ventricle, pumps the blood to the lungs. At the same time, the left ventricle pumps the blood to the rest of the body.

2.1.2 Heart Valves

There are four valves (tricuspid, pulmonary, mitral and aortic valves) in the human heart. These valves are unidirectional, maintaining the blood flow in one direction. The tricuspid valve is located between the right atrium and the right ventricle, while the pulmonary valve is positioned between the right ventricle and the pulmonary artery. The mitral valve is situated between the left atrium and left ventricle, and the aortic valve is between the left ventricle and the aorta. The tricuspid and mitral valves are called atrioventricular valves since they are between the atrium and the ventricle, whereas arterioventricular valves refer to the pulmonary and aortic valves as they are located between the artery and the ventricle. The arterioventricular valves are also called semilunar valves because of their half-moon-shaped leaflet. The atrioventricular valves are attached to the heart muscle (myocardium) by means of papillary muscles and fibrous cords. These valves are considered as active structures because it responds to myocardial contractions. The semi-lunar valves, on the other hand, do not have direct attachment to the myocardium, and therefore they are considered to function passively in response to blood flow.

These heart valves are susceptible to diseases and degeneration. And replacement of the diseased or defective valve is necessary. The valve that is most vulnerable to failure is the aortic valve. This is because of its higher operating pressure. As the results, there have been considerable studies concerning the aortic valve.

2.2 Aortic Valve

When the aortic valve opens, the blood flows into the aorta. When it closes, the blood is prevented from flowing back into the left ventricle. This opening and closing of the aortic valve is achieved by the movement of its three leaflets. Although the aortic valve performs the same task as the man-made valves, it is considered to be superior because no man-made valves, to date, have been able to work with the same efficiency and durability.

2.2.1 Aortic Valve Anatomy

Aortic valve is a “gate” between the left ventricle and the aorta. The arrangement of aortic wall with three aortic leaflets and three sinuses is called aortic root.

Figure 2.2 Location and geometry of the aortic valve

There are a few basic features of a common aortic root (see Figure 2.2).

1. Aortic Sinuses. As shown in the figure, the sinuses are three bulges that protrude from the aortic wall. Their role in the valve performance is to create the “back vortex” that assists the closing of the leaflets and to work together with the leaflet as a stress sharing structure.
2. The leaflet commissure. The commissure is the intersection region between the leaflets and the aortic wall. The tissue around this region is slightly thicker because the commissure acts as the stress concentration bearing region when the leaflet is pressurized by the back flow from the ascending aorta.

3. Leaflet coaptation area. This is the common meeting area of the aortic leaflets. A good coaptation area ensures that the leaflets can close properly and free from leakage and other sealing problems.
4. Leaflet belly. This is the lower part of the aortic leaflet. The belly is defined as the region that extends from the coaptation area to the leaflet attachment line, which is free to expand when the leaflets are pressurized.
5. Leaflet attachment line. This is an imaginary line that is used to define the leaflets attachment to the aortic wall. The line is formed from the intersection of the leaflets to the aortic wall.
6. Leaflet free edge. The top edge of the aortic leaflet facing the aorta.
7. Nodulus Arantii Region. The region at the free edge that lies on top of the coapting edges of the leaflets.

2.2.2 Aortic Valve Dysfunction

Several problems can interfere with the proper functioning of a valve. They may be congenital (abnormal from birth), infectious (endocarditic), inflammatory (rheumatic fever), or just wear and tear over the years.

In general, there are two major diseases with aortic valve. The first type is the valvular stenosis (Figure 2.3). Stenosis describes the condition when a valve does not open completely or the opening is too small, resulting in restricted blood flow. This abnormal structure causes turbulence in the blood flow that creates stress concentration points on the valve.

Figure 2.3 Aortic stenosis

(http://info.med.yale.edu/intmed/cardio/echo_atlas/entities/aortic_stenosis_senile.html)

The other disease is valvular regurgitation which is the failure of the aortic valve to close properly and hence allowing blood to leak back into the left ventricle, instead of continuing up through the ascending aorta. Over time the ventricle will be enlarged in order to pump more blood to compensate for the leakage.

In both cases the heart will have to work harder to maintain the same cardiac output. And this leads to enlargement of the heart and eventually fails. If the valve cannot be surgically repaired, the only alternative is to replace the defective valves with a mechanical or bioprosthetic valve.

2.2.3 Aortic Valve Replacement

The two main types of the artificial heart valve are:

- 1) Mechanical heart valve (MHV): Early MHV design is ball-in-cage type. Later, the caged disc, the tilting disc, and the bileaflet disc valves were introduced (Figure 2.4). Although the mechanical valves are very durable, they have risk of blood clots formation (thrombogenic) on valve components. Consequently, the patients receiving the mechanical valves must have constant anticoagulant intake for life. Currently the best-selling type of mechanical valve is the bileaflet design, which has demonstrated excellent durability and low thrombogenic rates.

Figure 2.4 Mechanical heart valves (www.edwards.com)

- 2) Bioprosthetic valve (BHV): The BHV may be constructed from tanned bovine pericardium, porcine aortic valves, or glycerol treated human tissues (Figure 2.5). They are not usually thrombogenic and thus have a significant advantage over the MHVs since the patients do not require long-term anticoagulant therapy (www.lerner.ccf.org). Because of their flexible properties, they demonstrate similar behavior with the natural aortic valve. The disadvantage of BHVs is apparent: the cells on the leaflets are no longer living tissue and therefore lacking of cellular regeneration. This makes them more vulnerable to damage. (Yoganathan, A.P., 1995)

Figure 2.5 Bioprosthetic heart valves (www.heartlab.robarts.ca)

2.3 Study of Failure in BHVs

The bioprosthetic valves (BHVs) usual failure is due to calcification and tearing damage. Large calcification growths can cause stenosis, and leaflet tearing can cause regurgitation (Vesely I., 2003). For many years, the calcification has been promoted as the main mechanism of valvular degeneration, with tearing damage as the outcome (Schoen FJ et al., 1987). But recently, researches have showed that mechanical damage and

calcification can be two separate processes (Schoen FJ et al., 1999). Actually the calcification is a relatively minor issue for BHVs failure. The majority of the damages occur strictly because of mechanical reasons (Vesely I et al., 2001). While it was initially believed that tensile stresses induced during the valve closure were responsible for the tearing, some studies have shown that tissue rupture is related to fatigue (Ishihara T et al., 1981; Vesely I et al., 1988, 2001) and the presence of the compressive stresses on the inner surface of the bent tissue (Krucinski S et al., 1993).

In order to develop more durable valves, the BHVs have been studied as a whole or divided into microstructures. The mechanical qualities of a tissue section and the flexural motions of a whole valve under the physiological condition have also been investigated. Besides experimental work, numerical methods were also introduced in the studies of the BHVs.

2.3.1 Experimental Study of BHVs

There are several aspects of the BHVs that have been observed. Several studies have focused on the BHVs material properties. They used uniaxial tests (Zioupos P et al., 1994; Hanlon JG et al., 1999), biaxial tests (Humphrey JD et al., 1987; Billiar KL et al., 2000), bending tests (Vesely I et al., 1989), and inflation tests (Zioupos P et al., 1992; Hanlon JG et al., 1999) to obtain the material properties. Their studies suggested that the current glutaraldehyde fixation process eliminated many of the beneficial, stress-reducing properties of the aortic leaflet. Additionally, some mechanical properties are still being disputed, e.g. whether bovine pericardium was isotropic or anisotropic. Furthermore, the test data have not been utilized to build more accurate numerical model.

Another group of researchers analyzed the failure of BHVs. The fatigue damage of the BHV tissue was discussed (Vyavahare N et al., 1998; Lee TC et al., 2001; Sacks MS et al., 1998, 2002). The studies suggested that these damage or de-naturalization of the tissue compositions would lead to the BHV deterioration.

To study the mechanical behavior of BHV tissue, leaflet specimen was usually cut to small sections in the traditional test. In order to analyze the valve behavior in intact status, photogrammetric technique was introduced. Photogrammetry is a process of making precise measurement by means of two-dimensional photographs. The main advantage of the photogrammetry is that it can measure the profile and motion of valve leaflet through images without contact with tissue. Chen L investigated the non-homogeneous deformation in the mitral valve leaflet with photogrammetric technique under a quasi-static pressure load (Chen L, 2004). In the investigation the effects of cellular contraction on aortic valve leaflet bending stiffness (Merryman WD et al., 2006), the photogrammetry was also used.

Photogrammetry can be applied not only in static but also in dynamic measurement. The quantification of heart valve leaflet dynamic deformation during the cardiac cycle is essential in understanding normal and pathological valvular functions, as well as in the design of the BHVs (Iyengar AKS et al., 2001). Vesely I first developed an isolated heart setup with a camera system to study the dynamic behavior of the porcine aortic valve (1990). Later, attempt to reconstruct the leaflet motion was carried out by Gao Z (1999), Gong MF (2001) and Iyengar AKS et al.(1999, 2001). Surface contour and curvatures at opening and closing phase were reconstructed. The information of the locations of the moving leaflet was carried out by using the structured light projected on the surface of leaflet. However, the structured light projection cannot follow the exact material point of the leaflet, thus strain cannot be obtained from these approaches. Hence the alternative techniques were carried out. Sack MS studied the cuspal motion by tracing the lead markers glued on leaflet through bi-planar X-ray (1999), while Gao ZB recorded the leaflet motion using a dual-camera stereo system (2000). The information on the location of the moving leaflet was mapped out using the markers on the leaflet surface. In addition, the surface geometry was used to compute change of curvature tensor.

A common conclusion from these studies suggested that future work should be extended to include reconstruction of a 3-D leaflet surface for numerical method.

2.3.2 Numerical Study of BHVs

Static Models

The direct measurement and observation of the aortic valve through experiment is still on-going. However, there is a lack of the universal technique that can provide a complete solution to the problems associated with the valve motion. As an alternative approach, the numerical method can help us to study the stress distribution in various BHVs models with different leaflet geometry and material properties. Using the numerical method, the causes of failure of BHVs can be further studied, the BHV mechanical properties can be evaluated, and finally an optimal BHV model can be designed (Hanlon JG et al., 1999).

In earlier studies, many researchers have studied the different aspect of the BHVs model. Sabbah et al. (1985) developed a finite element model of porcine bioprosthetic valve (PBV) by assuming that the geometry of a relaxed stent-mounted PBV was one half of an elliptic-paraboloid. They also considered the effect of a diseased PBV, which includes stiffening, focal calcium, and focal thinning. Hamid and his co-colleagues (1986) studied the influence of stent height on the closed BHV, concluding that the stress increases as the stent is getting shorter. Hancock heart valve prosthesis (one of BHV types) was studied by Rousseau et al. at 1988. The linear anisotropic viscoelastic behaviour of heart valve was modeled using membrane and truss element. Subsequently, in 1991 Chandran et al. focused their study on a polyurethane tri-leaflet heart valve in closed position, with rigid and flexible stents.

Huang and his co-colleagues (1990) developed a two-dimensional model to simulate leaflet flexure during valve opening. They analyzed the leaflet through two directional (radial and circumferential) sliced cross sections. In this study, the nonlinear material property of human pericardium was considered. The attachment to the stent was simulated in detail by progressive contact, and the coaptation of the leaflets was simulated by a straight contact line. They found that the regions of the valve that were

most heavily stressed were subjected to a strong bending, and the stress level depends on the boundary conditions and the initial configuration of the valve.

Krucinski et al (1993) used brick elements to analyze the different features of rigid, pliable, and expansible stent. The leaflets were assumed to be of uniform thickness, but with a non-linear elastic behavior. It was found that no substantial decrease in compressive stresses during the opening phase of the pliable stent with rigid stent. They concluded that high flexural and compressive stresses existed at the sites with high leaflet bending and might promote BHV failure. Although it was realized that the compressive stress during tissue bending was an important reason for the failure of BHVs, the bending was neglected in all these earlier numerical studies as the leaflet was regarded as tension-only structure.

To evaluate and optimize synthetic heart valves, Hart (1998) and Cacciola (2000) developed a three dimensional numerical model of a fiber-reinforced valve, which can alter the stress distribution to decrease the possibility of tearing and calcification. In the model, the material property of fiberless element was assumed to be incompressible and isotropic, and the fiber-cross element was designed with the homogenous orthotropic linear elastic properties. They compared the stress distribution of different number of fiber (1998), proving that the fiber took more than 60% of the maximum principal stresses.

Li et al. (1999, 2000) presented a nonlinear anisotropic model for porcine aortic heart valve. A nonlinear composite constitutive model was developed. The method was based on the microstructure of the leaflet and the linear constitutive model of the transversely isotropic materials. Comparing with the isotropic valve, it was found that the location of stress concentration for the anisotropic leaflet was different. The maximum longitudinal normal stress was increased, while the maximum transversal normal stress and in-plane shear stress were reduced.

Some earlier studies had shown that the hemodynamic performance of the valve was

almost the same whether it had a tubular root or a sinus root (Thubrikar MJ et al., 1997). Thus the sinuses might be only marginally important to the valve dynamics. To verify this assumption, Beck (2001) developed a model that compares the effect of the existence and the absence of the sinuses on the stresses in the leaflets. It was found that the stress in the leaflet belly was not affected by the geometry and stiffness of the root, but changed a lot along the attachment. From this study, the sinuses are believed to be important in minimizing the stress in the leaflet.

As the development of the artificial heart valve continues, the stentless valve models that share the stresses with aortic wall were studied (2000). Results showed that in the stentless models, the stresses were reduced by up to 75% (compared with the stented models with the same type of reinforcement).

In 1998, Grande et al. created an asymmetry finite element model from magnetic-resonance images of nine human valve-root specimens. This was the first time that the initial geometry of heart valve, including the leaflet and aortic wall, was developed from experimental measurement. This study explained the difference of the dilatation probability among the three sinuses. Furthermore, based on this model, some failures in the heart valves were studied, such as tissue change with age (1999) and aortic root dilatation (2000). In another study using valve-sparing techniques (2000), the sinuses were found to be significant for sparing the aortic valve. All the studies above only focused on the closed leaflet in a static analysis.

Though many researches attempted to understand the mechanics and the functions of the BHVs in different aspects, all these models used static finite element analysis with most simulations focused on the diastole state, i.e. when the aortic valve closes. But these were not enough to illustrate the whole process of the aortic valve movement in a cardiac cycle. Thus a dynamic simulation was needed. The displacements and stresses, which were associated with time, could be calculated by dynamic models.

Dynamic Models

In 1997, Thornton et al. presented a dynamic model for the stress analysis for a polypropylene leaflet. This was a bileaflet prosthetic heart valve. The shape of leaflet open state was modeled as its initial geometry. An isotropic linear elastic material was assumed, and shell elements with the capacity to include bending were used in the model. But in this simulation, only a small part of the cardiac cycle was used for the study, which did not show a complete period of the cardiac cycle.

Using the same geometry as Thornton's model, Patterson simulated the bioprosthetic heart valve with nonlinear material property in 1996. A strain energy density function, which was also described by Huang (1990) and Black (1991), was utilized in the nonlinear material model. The study showed that the nonlinear model was more responsive to the time-varying pressure wave. But, the Young's modulus, corresponding to the lower end of the nonlinear stress-strain characteristic, was much higher than the experimental data obtained.

Later in 1999, with the same strain energy density function, a dynamic model of a bioprosthetic porcine valve was developed by Chew et al. A closed prosthetic tri-leaflet valve with rigid stent was designed as the initial geometry. The model simulated the nonlinear elastic behavior of the fixed tissue and the reinforcement of the leaflets collagen with beam elements. Progressive damage was introduced into the models at a location where tears had been commonly found in vivo. This simulation showed that the rate of tearing increases with length.

The latest dynamic simulation of the aortic valve was modeled by Ramakrishna in 2002. The major aim of the study was to examine the leaflet-aortic root interaction during the cardiac cycle. In this model, an aortic valve in an opening state was assumed as the initial geometry. Starting from a particular stage in the diastolic cycle, a whole cardiac cycle was taken into account. However, it could not represent the incompressible property of

the biology tissue.

Arcidiacono investigated the leaflets performance by considering orthotropic mechanical characteristics (2005). The initial geometry was a closed PericarbonTM aortic tricuspid valve. Dynamic valve motion was simulated with a modified pressure difference between aorta and ventricle.

The valve opening and closing involves a strong interaction between blood and the surrounding tissue. In the numerical analysis so far, the leaflet was just moving under the assumed or standard uni-spatial pressure. It has been proven that the initial prototype and the geometry variations through time are very important to the stress analysis of BHVs (Gould et al. 1973). But, lack of detailed information on the interaction with fluid (blood) decreases the accuracy of the deformations obtained from the analysis. Several attempts have been made to analyze the valve behavior using numerical fluid-structure interaction models, e.g. see Peskin et al. (1995), Makhijani et al. (1997), Hart et al. (2000, 2002, 2003), and Carmody (2006). They tried to resolve the difficulty of the large deformation of the structural mesh that occurred in the fluid mesh. However, until now the fluid-structure simulation still could not present the real physiological environment. Thus the reliability of geometry variations simulated in these studies is still in question.

Conclusion

Based on the above literature review, it is known that the failures of the BHV are mostly due to the mechanical stress concentration, which occurs during systolic/diastolic phases. High stress is often generated in the regions undergoing high deformation. Finite element analysis (FEA) is widely used in studying the stress distribution in the BHVs. As the stress analysis is directly related with the geometry of the leaflet, it is necessary to ensure that the deformation of the leaflet is similar to what was observed during the experiment. From these numerical simulations, various profiles of the moving leaflet can then be obtained. However, the geometrical profiles of the folding and unfolding valve leaflet obtained from previous numerical simulation cannot represent the actual deformations of the valve leaflet. In the open literatures itself, there has been little information that relates

to the detailed time-spatial loads involving aortic valve leaflet under dynamic condition.

In order to simulate the actual deformations of BHVs, this PhD research attempts to include the reconstructed three dimensional leaflet contours into the FE analysis. A novel photogrammetric technique was used to capture and measure the dynamics of an aortic valve fixed in a mock physiologic flow loop of cardiac cycles. The 3-D displacements along the free-edge of the valve leaflet were then applied as the boundary conditions of the FE model. The ensuing numerical simulation would better mimic the actual aortic valve motion. The result shows that using these dynamic boundary displacement loads, the numerical model can complement the lack of fluid flow information and simulate the actual movement of the BHV more accurately.

Chapter 3 Experimental Methodology

3.1 Physiological Mock Flow Loop

The purpose of the physiological mock flow loop (PMFL) experiment was to investigate the motion of a bioprosthetic valve. The whole experimental setup is shown in Figure 3.1. The BHV was placed in the aortic position (A) in PFML. The PMFL was used to duplicate the function of the left side of the heart under physiological conditions. The PMFL system consists of three basic parts: the digitally controlled electromagnetic motor; the transparent aortic and ventricular valve testing chambers, and the adjustable windkessel afterload system that consists of two compliance units and two resistor units. The PMFL (Figure 3.2) was jointly built by the Cardiovascular Engineering Laboratory (University of Miami, USA) and Physiological Mechanics Laboratory (NTU).

Figure 3.1 Scheme of experimental system

(A: test aortic valve; AC: aortic conduit; CC: characteristic compliance; CR: characteristic resistance; EMM: electromagnetic motor; LA: left atrial chamber; M: mitral valve; P: piston; PA: aortic pressure probes; PC: peripheral compliance; PR: peripheral resistance; PV: ventricular

pressure probes; RO: rotometer; VR: variable resistance; VS: ventricular sac)

Figure 3.2 Physiological mock flow loop

3.1.1 Valve Testing Chamber

The valve-testing chamber includes a collapsible ventricular sac and an aortic conduit. The ventricle sac is fixed in the ventricle box. It divides the space in the box into two parts. The space outside the sac is filled with incompressible water and the space inside the sac is filled with glycerin solution. The sac is connected to the rest of the parts in the test rig via two holes, where the mitral valve and the aortic valve are mounted.

The pulsatile movement of the piston causes the water pressure in the ventricle box to change. The pressure acts on the ventricle sac and the glycerin solution in it. As a result, the solution is ejected out of the sac and flow through the whole circuit

cyclically, simulating the human heart beat.

The aortic conduit (Figure 3.3) is positioned at the downstream of the ventricular sac. The aortic conduit is an important part in the experiment because it is an interface between the PMFL and the optical measurement system. The aortic conduit has three symmetric sinuses at the base just like the human aorta. As shown in Figure 3.3, two inclining holes were drilled at the aortic conduit, through which borescopes (see section 3.2.1) can be inserted in to capture clear images of the BHVs. The angle between the two holes is about 32° . When inserted, the borescope seals the hole so that no solution will leak out and the flow field is minimally affected.

Figure 3.3 Aortic conduit

(the holes when no borescopes insert in are blocked up with two delrin sticks)

Two physiologic pressure transducers were installed in the PMFL. The ventricular pressure port was located in the ventricular sac about 10mm upstream of the aortic valve, and the aortic pressure port was located in the aortic conduit 90mm downstream of the aortic valve.

During the test, analog pressure signals were acquired by a DAP board (DAP 800 Data Acquisition Processor; Microstar Laboratories, Inc., Bellevue, WA) with a 4-channel 12-bit analog-to-digital converter. The sampling time of the signals was $40\ \mu\text{sec}$. For each channel, the acquired data were filtered at the DAP by averaging every group of 20 successive data points to give 1 data point. Thus the time interval between two consecutive data points of each output waveforms is 3.2 ms.

3.1.2 Windkessel Afterload

The afterload system includes two compliance chambers and two linear resistances. After aortic conduit, fluid flows into the characteristic compliance chamber, and characteristic resistance. The flow continues through a peripheral compliance and a peripheral resistance. The resistance can be tuned continually while PMFL is operating. This enables the system to achieve the physiologic pressure and flow waveforms required for valve testing.

After the aortic and mitral valves had been installed in the PFML, the flow loop was started. The systolic/diastolic aortic pressure levels were maintained at about 120/80 mmHg throughout the experiments. The heartbeat was set to 70 beats/min. The flow rate, about 4~5 liter/min, was measured by a calibrated rotameter at the upstream of the atrium chamber. The experiment was performed with the ambient temperature range 23°C to 25°C . The testing fluid was 33.3% water-glycerin solution, which at room temperature exhibits viscosity and density similar to human blood under physiological conditions.

Figure 3.4 illustrates the pressure waveform obtained from this PMFL test. Comparing with physiological values, the aortic pressure is a bit higher in systolic stage and lower in diastolic stage. Physiologically, common human aorta is made

from elastic tissue that will cushion the pulse pressure. The expansion of human aorta will reduce pressure at the systolic stage and the contraction will increase the aortic pressure at the diastolic stage. However, because of the requirement to introduce the camera in the PMFL, a rigid aortic conduit was used in this test.

Figure 3.4 Ventricular and aortic pressure waveforms

3.2 Image Capturing

3.2.1 Borescope

Borescope is a hollow long thin stainless-steel tube in which lenses are installed. It can be inserted into small areas where a normal optical camera head cannot reach. The borescopes were selected to record the motion of valve based on two reasons. First, it can be used to focus on one valve leaflet in a relatively close distance, so that detailed observation can be made. Second, the insertion of the borescope into the aortic conduit does not affect the fluid flow through the PMFL significantly. (Gong MF, 2001)

Two rigid borescopes were used in this test. The borescopes used are Olympus series 5 borescopes (Figure 3.5), with 8 mm diameter and 240 mm length of working tube. Since the shutter speed is high, sufficient brightness is required. Light power Olympus ILK-5 was used to supply illumination to borescopes through light guide cables.

Figure 3.5 Light power, borescope, light guide cable and C-mount lens

3.2.2 CCD Camera

Two high-speed CCD cameras were employed in the experiment. One is FASTCAM-Super (KODAK Motion Corder Analyzer, SR series, Japan), and the other one is FASTCAM-PCI R2 model 500 (Japan) (see Figure 3.6). The FASTCAM-Super has several components: a portable processor unit, a power supply, a camera head and a viewer. The FASTCAM-PCI R2 only has a camera head and a grabber board (PCI board), which needs to be installed in an ATX standard PCI slot of a PC. With FASTCAM control software (Photron FASTCAM Viewer 2.2.2.1), images can be viewed and recorded into the PC.

FASTCAM-Super

FASTCAM-PCI R2

Figure 3.6 High speed cameras

In the subsequent sections of this PhD thesis ‘camera-1’ refers to FASTCAM-Super and camera-2 refers to FASTCAM-PCI R2.

Camera-1 was controlled through push-button on the processor unit, and camera-2 was controlled through the software Photron FASTCAM Viewer installed in PC. In term of functionality, these two cameras are the same. Both cameras are designed to work at 250 frame/sec with 512×480 pixels or 500 frame/sec with 512×240 pixels. In order to obtain larger image, the 250frame/sec (fps) was used. Using this image size, the memory of camera-1 can save up to 546 frames. Because of a larger memory, camera-2 can save more frames. However, in order to pair the two camera images, only 546 frames were used in this experiment. Since the heart beat rate was set to 70 beat/min, these 546 frames covered over 2.5 cardiac cycles.

The camera’s processor treats memory as a circular buffer. When the trigger signal is received, pictures are stored in sequence with the current one replacing the oldest one in memory. Because the cameras were used to capture a series of motion, the way to trigger them was important. The cameras have four trigger modes: start, records 2.2 seconds after receiving trigger signal; center, records 1.1 seconds before trigger and 1.1 seconds after trigger; end, records 2.2 seconds before trigger; and random, records one, ten, or fifty frames for each trigger signal received by the processor. In this experiment, the start trigger mode was chosen to capture the moving leaflet. The two cameras were triggered synchronously by one start-up signal.

During one cardiac cycle of 0.857sec, the period of leaflet opening stage is about 0.3sec, and the transition period from close position to full open position is only 0.03sec. Since the bioprosthetic aortic valve moves relatively fast during the cardiac cycles, the captured images will be blurred if the camera exposure time is relatively long. Thus the shutter speed was controlled. When the cameras were working at 250frame/sec, the shutter speed could vary from $1/250 \text{ sec}^{-1}$ to $1/10,000 \text{ sec}^{-1}$. Shutter speed of $1/1000 \text{ sec}^{-1}$, $1/2000 \text{ sec}^{-1}$, and $1/3000 \text{ sec}^{-1}$ were tested. The later two shutter

speeds enhanced the images definition, but the illumination became relatively poor when the shutter speed was $1/3000\text{sec}^{-1}$. Thus the shutter speed was set to $1/2000\text{sec}^{-1}$. Images could be well identified with little blurring found at this shutter speed.

Each CCD camera head was mounted on borescope through a C-mount adaptor (C-mount Olympus AK2-10C). The images captured by camera-1 were downloaded into PC through an Adaptec SCSI PCI Card. The images captured by camera-2 were saved in PC using software Photron FASTCAM Viewer. The experimental results might be affected by the lens distortions. Hence, before recording the leaflet motion, the camera setup (including the camera head, C-mount adaptor and borescope) were examined.

3.3 Photogrammetry

3.3.1 Basic Principle

In order to follow the position of a certain material point, the dual-camera stereo photogrammetry was used. Using this technique, an actual motion of the free-edge in the aortic valve leaflet was obtained.

First, the two high-speed CCD cameras captured the images of the same BHV leaflet simultaneously during several steady cardiac cycles. The deformations of the leaflet during the cardiac cycle could be expressed by the 3-D coordinate changes of the marker matrix on the leaflet surface (see Figure 3.7). The attached markers were made from ink drawn on the surface. The difficulty with the ink marker is that it always tends to seep into a larger area when the leaflet surface is humid. Thus, the leaflet surface of a porcine BHV was kept in the air to damp-dry first. Then, some black markers were applied on the surface manually with a permanent fast dry extra-fine pen (Twin Marker, Pilot, Japan). When the markers were almost dry, the porcine valve

was stored in the (glycerin) solution to avoid leaflet dysfunction due to over-dryness. The marker sizes are about 0.2mm to 0.4mm. There are almost 120 dots marked on the surface with 21 markers situated along the free-edge. The inter-spacing is about 0.9mm to 1.7mm. Though it was relatively difficult to catch the whole markers by the two cameras during the whole period, at least all the markers along the free-edge were captured clearly. A centric search algorithm was then designed to determine the markers location.

Figure 3.7 Tested bioprosthetic valve

Secondly, after recording the aortic valve motion under the pulsatile loading, a relaxed valve figure and a short cylinder (see Figure 3.8), which are used as the calibration structures, were captured without changing the camera configurations. The calibration cylinder had a diameter of 12mm and a height of 8mm. The 85 markers were machined on the white delrin cylinder; 55 were on the top circular surface and 30 were on the side face. The size of each dot was about 0.5mm to 0.8mm. The inter-spacing between dots was from 0.7mm to 2mm. The world coordinate system was originated on the center markers of the cylinder, as shown in Figure 3.8. The 3-D locations of the valve were reconstructed based on this coordinate system.

Figure 3.8 World coordinate system

Thirdly, the 3-D coordinates of the markers on the leaflet free-edge were calculated from the pairs of instantaneous digital images using Standard Direct Linear Transformation (DLT) method. The DLT parameters were first calculated through the known locations of calibration cylinder markers. Then, the marker locations on the leaflet free-edge could be retrieved from paired images through these parameters. The 3-D coordinates of each marker could provide leaflet displacement during a whole cardiac cycle.

Though the two cameras were synchronized by a trigger, the DLT method (Philippe Pourcelot et al., 2000) further ensured temporal precision of the 3-D coordinates calculation. The displacements of the free-edge markers would be used as moving boundary conditions on the dynamic numerical model, which will be described in later chapter.

3.3.2 Direct Linear Transformation

Direct Linear Transformation (DLT) is a common approach used in biomechanics to reconstruct 3-D space from two or more 2-D images. DLT was first carried out by Abdel-Aziz and his colleagues (1971) and developed by Marzan et al. (1975). In brief, initially paired images from a reference structure, containing known markers in 3-D

space, are obtained. This structure was then removed and the test object was captured in the same space with the same camera set-up. The 2-D images of both the reference structure and object were then digitized and the unknown 3-D coordinates of each of the object's markers were determined by solving a set of DLT basic equations. (Wood GA et al., 1986)

The primary advantages of this technique are: (i) the optical axes of the cameras do not need to intersect; (ii) the positions of the cameras are arbitrary and do not need to be measured; (iii) only two cameras are required; and (iv) additional cameras can be accommodated. (Challis JH et al., 1992)

For simplicity, a camera can be modeled by a pin-hole camera model (Figure 3.9). Recording images using a camera is equivalent to mapping object point in the object space to image point in the image space. Object M is mapped directly to the projected image m through a projection center C. The projected plane is called image plane. The line drawn from the projection center C and perpendicular to the image plane is called the principal axis. The principal point c is the intersection of the principal axis with the image plane.

Figure 3.9 Pin-hole camera model and four coordinate systems

In total, there are four coordinate systems that are used to describe the transformation between the object space and the image space. They are (i) Pixel coordinate system

(u,v), which represents the image using the column and row index of the image array; (ii) Image plane coordinate system (x,y), which uses true dimension instead of array index to represent the image. Its origin is the principal point c. The x-axis is in the direction of increasing column index and the y-axis is in the direction of increasing row index; (iii) Camera coordinate system (Xc,Yc,Zc), is the coordinate system located at the projection center C with Xc- and Yc-axes parallel to the x and y axes in the image plane, and the Zc axis, also called optical axis, perpendicular to the image plane; (iv) World coordinate system (Xw,Yw,Zw), which is the Cartesian system established to represent points on the object. For the ease of coordinate conversion, all of the spatial coordinate systems are right-handed systems.

Conversion from world coordinate system to pixel coordinate system includes three transformations: from the world coordinate system to the camera coordinate system, from the camera coordinate system to the image plane coordinate system, and from the image plane coordinate system to pixel coordinate system.

In the Cartesian coordinate system, the perspective transformation is nonlinear. For ease in calculation, the transformation can be linearized by mapping the Cartesian coordinates into homogeneous coordinates. A point M with Cartesian coordinates (X,Y,Z) has the homogeneous coordinates:

$$\tilde{\mathbf{M}} \cong \begin{bmatrix} X \\ Y \\ Z \\ 1 \end{bmatrix} \quad \text{or} \quad \mathbf{M} = k \begin{bmatrix} X \\ Y \\ Z \\ 1 \end{bmatrix} = \begin{bmatrix} kX \\ kY \\ kZ \\ k \end{bmatrix}$$

where “ $\cong$ ” means equality up to a scale factor.

From world coordinate system to camera coordinate system:

It is obvious that the transformation between these two coordinates systems is rigid body motion. Therefore, the coordinates for each point at the world coordinate system

are transformed into their coordinates in the camera coordinate system by rotation and translation:

$$\mathbf{X}_C = \mathbf{R}\mathbf{X}_W + \mathbf{T} \quad (3.1)$$

where $\mathbf{R}$ is a 3 by 3 matrix for rotation and $\mathbf{T}([X_0, Y_0, Z_0]^T)$ is the position of the origin of world coordinate system with respect to the camera coordinate system.

If the rotation of the coordinate system can be specified by three angles: rotation α about the x-axis, rotation β about the new y-axis, and rotation γ about the z-axis, each element of the rotation matrix $\mathbf{R}$

$$\mathbf{R} = \begin{bmatrix} R_{11} & R_{12} & R_{13} \\ R_{21} & R_{22} & R_{23} \\ R_{31} & R_{32} & R_{33} \end{bmatrix} \quad (3.2)$$

can be expressed as follows:

$$\begin{aligned} R_{11} &= \cos \beta \cos \gamma \\ R_{12} &= \cos \beta \sin \gamma \\ R_{13} &= -\sin \beta \\ R_{21} &= \sin \alpha \sin \beta \cos \gamma - \cos \alpha \sin \gamma \\ R_{22} &= \sin \alpha \sin \beta \sin \gamma + \cos \alpha \cos \gamma \\ R_{23} &= \sin \alpha \cos \beta \\ R_{31} &= \cos \alpha \sin \beta \cos \gamma + \sin \alpha \sin \gamma \\ R_{32} &= \cos \alpha \sin \beta \sin \gamma - \sin \alpha \cos \gamma \\ R_{33} &= \cos \alpha \cos \beta \end{aligned}$$

The transformation, expressed in homogeneous coordinates is

$$\tilde{\mathbf{X}}_C \cong \begin{bmatrix} \mathbf{R} & \mathbf{T} \\ 0_3^T & 1 \end{bmatrix} \tilde{\mathbf{X}}_W \quad (3.3)$$

From camera coordinate system to image coordinate system:

For a pinhole model, the focal length f is the distance between the origins of camera and image coordinate systems (C_c , see Figure 3.9). Applying similarity theorem, the perspective projection formula can be obtained as

$$\begin{cases} x = f \frac{X_c}{Z_c} \\ y = f \frac{Y_c}{Z_c} \end{cases} \quad (3.4)$$

It can be further expressed in homogeneous coordinates as:

$$\tilde{\mathbf{x}} \equiv \begin{bmatrix} x_1 \\ x_2 \\ x_3 \end{bmatrix} \equiv \begin{bmatrix} f & 0 & 0 & 0 \\ 0 & f & 0 & 0 \\ 0 & 0 & 1 & 0 \end{bmatrix} \begin{bmatrix} X_c \\ Y_c \\ Z_c \\ 1 \end{bmatrix} \equiv \begin{bmatrix} f & 0 & 0 & 0 \\ 0 & f & 0 & 0 \\ 0 & 0 & 1 & 0 \end{bmatrix} \tilde{\mathbf{X}}_c \quad (3.5)$$

where $x = \frac{x_1}{x_3}, y = \frac{x_2}{x_3}$.

From image coordinate system to pixel coordinate system:

The coordinates of the principal point c in the pixel coordinate system are $[u_0, v_0]^T$ (see Figure 3.10). Let S_x and S_y be the effective sizes of the pixel (mm) in the horizontal and vertical direction. Then, the transformation is basically the image coordinates scaled by S_x, S_y and translated by $[u_0, v_0]^T$.

$$\begin{cases} u = \frac{1}{S_x} x + u_0 \\ v = \frac{1}{S_y} y + v_0 \end{cases} \quad (3.6)$$

Figure 3.10 Image coordinate system (c,x,y) and pixel coordinate system (o,u,v)

It is expressed in homogeneous coordinates as:

$$\tilde{\mathbf{u}} \equiv \begin{bmatrix} u_1 \\ u_2 \\ u_3 \end{bmatrix} \equiv \begin{bmatrix} 1/S_x & 0 & u_0 \\ 0 & 1/S_y & v_0 \\ 0 & 0 & 1 \end{bmatrix} \tilde{\mathbf{x}} \quad (3.7)$$

where $u = \frac{u_1}{u_3}, v = \frac{u_2}{u_3}$.

The expressions among the homogenous coordinates are linear. By assembling the expressions (3.3), (3.5), and (3.7), the transformation between the world coordinate system and the pixel coordinate system can be obtained as:

$$\begin{aligned} \tilde{\mathbf{u}} \equiv \begin{bmatrix} u_1 \\ u_2 \\ u_3 \end{bmatrix} &\equiv \begin{bmatrix} 1/S_x & 0 & u_0 \\ 0 & 1/S_y & v_0 \\ 0 & 0 & 1 \end{bmatrix} \begin{bmatrix} f & 0 & 0 & 0 \\ 0 & f & 0 & 0 \\ 0 & 0 & 1 & 0 \end{bmatrix} \begin{bmatrix} \mathbf{R} & \mathbf{T} \\ \mathbf{0}_3^T & 1 \end{bmatrix} \tilde{\mathbf{x}}_w \\ &\equiv \begin{bmatrix} f/S_x & 0 & u_0 & 0 \\ 0 & f/S_y & v_0 & 0 \\ 0 & 0 & 1 & 0 \end{bmatrix} \begin{bmatrix} \mathbf{R} & \mathbf{T} \\ \mathbf{0}_3^T & 1 \end{bmatrix} \tilde{\mathbf{x}}_w \equiv \begin{bmatrix} P_{11} & P_{12} & P_{13} & P_{14} \\ P_{21} & P_{22} & P_{23} & P_{24} \\ P_{31} & P_{32} & P_{33} & P_{34} \end{bmatrix} \begin{bmatrix} X_w \\ Y_w \\ Z_w \\ 1 \end{bmatrix} \end{aligned} \quad (3.8)$$

where

$$\begin{aligned} u &= \frac{u_1}{u_3} = \frac{P_{11}X_w + P_{12}Y_w + P_{13}Z_w + P_{14}}{P_{31}X_w + P_{32}Y_w + P_{33}Z_w + P_{34}} \\ v &= \frac{u_2}{u_3} = \frac{P_{21}X_w + P_{22}Y_w + P_{23}Z_w + P_{24}}{P_{31}X_w + P_{32}Y_w + P_{33}Z_w + P_{34}} \end{aligned}$$

Since u and v are basically proportion between u_i ($i = 1,2,3$), all the coefficients P_{ij} can be divided by P_{34} without changing the values of u , v . By rearranging the coefficients P_{ij} , the standard 3-D DLT equations can be derived as:

$$\begin{aligned} u &= \frac{L_1 X_W + L_2 Y_W + L_3 Z_W + L_4}{L_9 X_W + L_{10} Y_W + L_{11} Z_W + 1} \\ v &= \frac{L_5 X_W + L_6 Y_W + L_7 Z_W + L_8}{L_9 X_W + L_{10} Y_W + L_{11} Z_W + 1} \end{aligned} \quad (3.9)$$

where,

$$[f_x, f_y] \equiv \left[\frac{f}{S_x}, \frac{f}{S_y} \right]$$

$$\begin{aligned} L_1 &= (f_x R_{11} + u_0 R_{31}) / Z_0 \\ L_2 &= (f_x R_{12} + u_0 R_{32}) / Z_0 \\ L_3 &= (f_x R_{13} + u_0 R_{33}) / Z_0 \\ L_4 &= (f_x X_0 + u_0 Z_0) / Z_0 \\ L_5 &= (f_y R_{21} + v_0 R_{31}) / Z_0 \\ L_6 &= (f_y R_{22} + v_0 R_{32}) / Z_0 \\ L_7 &= (f_y R_{23} + v_0 R_{33}) / Z_0 \\ L_8 &= (f_y Y_0 + v_0 Z_0) / Z_0 \\ L_9 &= R_{31} / Z_0 \\ L_{10} &= R_{32} / Z_0 \\ L_{11} &= R_{33} / Z_0 \end{aligned}$$

Coefficients L_1 to L_{11} are the DLT parameters that reflect the relationship between the world coordinate system and the pixel coordinate system. Equation (3.9) can be rewritten as:

$$\begin{bmatrix} X_i & Y_i & Z_i & 1 & 0 & 0 & 0 & 0 & -u_i X_i & -u_i Y_i & -u_i Z_i \\ 0 & 0 & 0 & 0 & X_i & Y_i & Z_i & 1 & -v_i X_i & -v_i Y_i & -v_i Z_i \end{bmatrix} \begin{bmatrix} L_1 \\ L_2 \\ L_3 \\ L_4 \\ L_5 \\ L_6 \\ L_7 \\ L_8 \\ L_9 \\ L_{10} \\ L_{11} \end{bmatrix} = \begin{bmatrix} u_i \\ v_i \end{bmatrix} \quad (3.10)$$

The equation (3.10) shows that each world-pixel pair generates 2 equations. In order to obtain all the eleven DLT coefficients, at least 6 pairs of known world-pixel coordinate points are needed. Least Squares technique was applied to solve for the DLT coefficients of each camera.

Equations (3.10) can be further expressed as:

$$\begin{bmatrix} L_1 - uL_9 & L_2 - uL_{10} & L_3 - uL_{11} \\ L_5 - vL_9 & L_6 - vL_{10} & L_7 - vL_{11} \end{bmatrix} \begin{bmatrix} X \\ Y \\ Z \end{bmatrix} = \begin{bmatrix} u - L_4 \\ v - L_8 \end{bmatrix} \quad (3.11)$$

Based on equation (3.11) at least two cameras should be used in order to obtain the 3-D coordinates. After L_1 to L_{11} have been worked out, the 3-D locations of other objects, which are captured with the same camera set-up, can be solved from equation (3.11) also using the Least Squares technique.

More on DLT method

The DLT method is commonly used in kinematics analysis of human and animal movement. This is due to the accuracy of the results obtained and the great flexibility

in camera set-up. Since its development in 1971, a number of researchers have examined and improved the method. The traditional DLT has 11 coefficients, while 12 and 16 coefficients were used in the refined DLT method (Abdel-Aziz YI et al., 1971; Marzan GT et al., 1975; Hatze F, 1988). However, 12 coefficients DLT method was proved to have no apparent benefit as compared with the traditional one. The 16 coefficients DLT may improve the results under certain conditions, but for ease of application, the researchers still utilized the traditional DLT method in later literatures. Previous works show that DLT method can give good reconstruction accuracy when the unknown object location is inside the scope of calibration structure (Wood GA et al., 1986; Hatze H, 1998; Challis JH et al., 1992; Gazzani F, 1993). The longer the distance calibration structure, the lower the reconstruction accuracy. Later Hinrichs RN showed that the reconstruction would be accurate if the level of extrapolation is less than 100% of the calibration volume (1995). The distribution of the calibration points would also affect the calculation results (Challis JH et al., 1992; Chen L et al., 1994). The calibration points with even distribution are needed. In practice, a camera setup with a base ratio, ranging from 1:3 to 2:1, appears to be equally reliable (Putnam C, 1979; Neal R, 1983; Wood GA et al., 1986). The base ratio is the ratio of the perpendicular distance (from a mid-camera point to the test object) *and* the distance between the two cameras. In the current experiment, the ratio was ~ 2:1.

Since the cameras captured the tested aortic valve inside the glycerin solution through the borescopes, the refraction between the air and the solution should be considered. Kwon developed a localized calibration method in underwater motion analysis (2000). The whole volume was divided into several sub-volumes. Calibration coefficients (L_1 — L_{11}) of sub-volume were used to reconstruct the world coordinates of marker that belongs to the sub-volume. However, potential problem with this method is that the space discontinuity due to switching of the sub-volume/area from one to another, and this problem is hard to avoid. Previous research has shown that the effect of refraction was related with the camera-to-interface distance and the interface-to-calibration-frame distance (Kwon YH, 1999). The longer these distances,

the smaller the reconstruction errors. In the localized calibration method, sub-volume instead whole volume was used to increase the reconstruction precision, because camera-to-interface distance and the interface-to-calibration-frame distance elongated by using the sub-volume (Kwon YH, 2000). In the current experiment, these two distances is relatively large as compared to the aortic valve size. This is especially true for the camera-to-interface distance (CCD to the head of the borescope; about 40cm long). Thus, the refraction error was made negligible.

The DLT method was implemented through KinMat, a tool box for 3-D kinematic analyses by Christopher R and Ton van Den B. This tool box is web-free software. Several valuation tests were done before applying this method.

3.3.3 Cameras synchronization

As the aortic valve moves at a relatively very high speed, a small time offset between two cameras may result in high inaccuracy. Though two high-speed cameras were synchronized by using one trigger signal, an extra effort was carried out to minimize the time offset. First, a numerical method was designed to test the synchronization quality. The method enables the synchronization of multiple cameras by exploiting several aspects of the DLT method mentioned in Philippe P et al., 2000. The method was used to estimate the time offset (phase difference) between the cameras, and then the offset was used to further synchronize the recordings of the cameras.

Equation (3.9) is the standard 3-D DLT equation. One may include the optical errors as:

$$\begin{aligned} u + \delta u &= \frac{L_1 X_w + L_2 Y_w + L_3 Z_w + L_4}{L_9 X_w + L_{10} Y_w + L_{11} Z_w + 1} \\ v + \delta v &= \frac{L_5 X_w + L_6 Y_w + L_7 Z_w + L_8}{L_9 X_w + L_{10} Y_w + L_{11} Z_w + 1} \end{aligned} \quad (3.11)$$

where $(\delta u, \delta v)$ are the errors associated with the coordinates.

After the Least Squares technique has been applied to equation (3.10) to solve the DLT coefficients, the DLT errors can be obtained from equation (3.11) for each camera.

One camera was chosen as the reference, and the other camera was synchronized with it. Although the two cameras were triggered by the same start signal, the second camera may be earlier or later than the reference camera. The phase (time) difference between the two cameras can be estimated by searching for the time difference value which minimizes the mean DLT errors $(\delta u, \delta v)$ of a moving point (Philippe Pourcelot et al., 2000). The mean DLT error per frame and per camera can be calculated as follows:

$$m_{err} = \frac{\sum_{j=2} \sum_{i=1}^{NbF} \sqrt{\delta u_{i,j}^2 + \delta v_{i,j}^2}}{2NbF} \quad (3.12)$$

where $\delta u_{i,j}$ and $\delta v_{i,j}$ are the DLT errors calculated at frame i on camera j along the camera image u and v axis, respectively, and NbF is the total number of frames.

The phase difference was found to be within one frame period (0.004s). This value was then used to synchronize the images captured by the cameras.

Chapter 4 Experimental Result

4.1 Camera Distortion

In Chapter 3, the experimental procedure using an ideal camera model (pin-hole model) has been described. However, the cameras used in the experiments are not perfect. Some distortions in the captured images were noted. It is important to minimized distortion so that reliable result can be obtained. Hence, the camera setup was investigated using a total of 20 images of planar checkerboard (see Figure 4.1), which were processed through Matlab camera calibration toolbox.

Figure 4.1 20 images of a planar checkerboard

The camera setup in this calibration included camera head, C-mount adaptor, and borescope. A printed square frame was pasted on a stainless steel plane to function as the planar checkerboard. The length of each square was 1mm. The checkerboard was

then placed at 20 different positions and was recorded. With these images, the distortion of the camera setup was calculated through the calibration toolbox. In this investigation the corners and grids in each image were extracted as the known parameters for analysis. Figure 4.2 shows the complete distortion model of the camera setup. It was found that the principle point (refer to Figure 3.9) was deviated from the center of images. There are two reasons for this finding. Firstly, the image center of camera does not coincide with the optical axis. Secondly, the camera, C-mount, and borescope are not coaxial. The distortion was large at the right and the bottom sides of images. To get reliable images, the test objects would be captured in the low distortion region.

Figure 4.2 Lens distortions in the camera set-up

4.2 Direct Linear Transform Test

Two 3-D reconstruction tests were developed to assess the experimental techniques developed in Chapter 3. In the first test, the points in the same object structure were reconstructed. In the second test, a delrin laddertron was used for reconstruction of a ruler.

4.2.1 Test 1

The test object was three square frames attached on three perpendicular surfaces. The grid size of the top frame was 0.5mm and the grid size of two side frames was 1mm. The structure was placed under water, and was observed by two cameras with borescopes inserted in the water. The vertex of the three surfaces was appointed as the origin of world coordinate system (Figure 4.3).

Figure 4.3 Grid frames captured by two cameras

For this test, the reference points and the extracted points were in the same structure. Referring to Figure 4.4, ten grid corner points were chosen as the calibrated reference points (*). The developed method was then used to reconstruct the rest of the points. The reconstructed points were then compared with known original coordinates. The maximum error for these reconstructed points was 0.3225mm, and the mean error is 0.0859mm. Comparing with the characteristic size (the diagonal of the calculated structure) of about 13.8mm, the error was acceptable (the maximum error was 2.3% and the mean error was 0.6%; both were less than 5%).

Figure 4.4 Reconstructed 3-D coordinates of points

(o: original coordinates; $\therefore$: reconstructed coordinates; *: 10 reference points)

4.2.2 Test 2

A delrin laddertron and a ruler (see Figure 4.5) were used in this second test. The radius of the top surface of the laddertron was 9.5mm, and the bottom surface radius was 14mm. The lines were drawn carefully every 10° across the top and bottom surface respectively. 45° lines were also drawn. A ruler was placed right under the laddertron (touching the bottom surface of the laddertron). The world coordinate system was based on the bottom surface center. X axis referred to the first line direction, and y axis referred to the anti-clockwise 90° line direction. Reference point for the world coordinates could be calculated from the radius and angle.

In this test, different number of reference points was used. First, 6 points were chosen (see Figure 4.5 (a)), and then the reference point number was increased to 15 (see Figure 4.5 (b)). The distance between the 11th and 12th scale on the ruler were calculated using the developed method. With 6 reference points, reconstruction of the

distance between these two scales was 10.33mm with an error of 3.3%. With 15 reference points, the calculated distance between two scales was 10.03mm, with an error of 0.3%. This study shows that the more the reference points are, the more accurate the result obtained. This test also incorporated the 3-D reconstruction of points outside the reference structure. The extrapolated object (ruler) was quite close to the reference structure (less than 50% of the characteristic reference-structure size). The results showed that the accuracy is acceptable.

Figure 4.5 Reference laddertron and ruler

4.3 Images of Bioprosthetic Valve Motion

After the camera setup and the developed method tested, the bioprosthetic test valve was then mounted on the aortic valve position in the PMFL. Because the cameras could not capture the whole aortic valve, the marked leaflet was placed in such a way that camera could capture the whole free edge during the whole cardiac cycle.

After PMFL function steadily with aortic pressure of around 120/70, the cameras were triggered to start recording the leaflet motion. The corresponding pressure curves were saved. Images were captured under the shutter speed of $1/2000\text{sec}^{-1}$ (the optimum shutter speed). Figure 4.6 was part of the paired images captured during a typical cardiac cycle.

relaxed valve

Figure 4.6 Paired images of leaflet motion during a cardiac cycle

Generally, numerical model (finite element model) needs a stress-free initial geometry. Apparently the leaflet configuration under the pulsatile pressure was not suitable for the finite element model because it was not stress-free. Thus, the leaflet configuration under a much low pressure (about 3cmH₂O) was recorded to obtain a relaxed leaflet-geometry.

The calibrated delrin cylinder was put in the aortic conduit, standing on the aortic valve perpendicularly (see Figure 4.7). The world coordinate system had been selected (refer to Figure 3.8). Next, all the points on the calibrated structure were used as reference points. Their locations could be obtained from known radius and angle.

Figure 4.7 Delrin cylinder as the calibrated structure

The images of the BHV showed that most of time the leaflet closed and little changes were observed. From 214 frames in one cardiac cycle, only 95 frames (# 156 to # 250) that showed significant movement. Hence, only those 95 frames were processed for the later reconstruction.

At the start of systole, the tightly stretches valve relaxed with the increased ventricular pressure. When the ventricular pressure caught up with aortic pressure, the aortic valve opened (frame # 164). The images showed that the leaflet heaved first in lower part of coaptation area and belly area, and then the free edge follows. In about 0.028sec the valve moved from the closed position to the largest open position (frame # 171). After that, the valve swung under the ejected flow with a relatively complex deformation (until frame # 221). From frame # 221 to # 236, the valve experienced a slow closing motion. At last, the valve closed sharply and quickly (From frame # 236 to # 239). The lower part of the coaptation area was closed first, and the upper part was dragged to a closed position later. The leaflet bumped with each other in a short time and then stabilized under the large pressure in diastole.

Results showed that the behaviors of the leaflet under the pulsatile loading were complicated. Previous studies used the images at different temporal points during different cardiac cycles based on the assumption that the motion of the leaflet had excellent cycle-to-cycle repeatability (Gao ZB et al., 2000; Iyengar AKS et al., 2001). Actually, by comparing the valve deformation in different cardiac cycles of this experiment, the leaflet motions were not identical except for the closed leaflet configuration. The images of the slow closing motion were chosen for this comparison (Figure 4.8) since the slight temporal difference would not affect the

observation significantly. It is clear that processing the images in one cardiac cycle is better than picking them from different cycles.

Figure 4.8 Comparison of deformation in the same time at different cardiac cycles

4.4 3-D coordinate Reconstruction of the Free Edge

After the images were recorded, they were used to reconstruct the 3-D coordinates in the world coordinate system. From the images of calibrated points on delrin cylinder and the known 3-D coordinates of these points, the DLT coefficients of each camera set were calculated and then used to reconstruct the 3-D coordinates of leaflet free edge markers.

Numerical synchronization of two cameras (as described in section 3.3.3) was also carried out. The value of the phase difference corresponding to the lowest mean DLT error was determined using an iterative method (Pourcelot P, 2000). All the 21 markers along free edge were used in the calculation. Since the phase difference corresponding to the minimum mean-DLT-error was slightly different among the 21 markers, an average value was used. Mean DLT error versus phase difference between two cameras is shown in Figure 4.9. The phase difference value between the two cameras was about 0.33 of time period or 0.004ms; the right camera actually captured images of leaflet motion slightly later than the left camera.

Figure 4.9 Mean DLT error versus phase difference between two cameras
(For clarity, only 6 markers results are drawn above)

The phase difference value was then used to synchronize both recordings. From the synchronized 2-D coordinates of paired images, the 3-D coordinates of the free-edge were reconstructed (Figure 4.10).

Since the tested valve was a stented type, the attachment line (refer to Figure 2.2) was assumed to be rigid. The commissures are important points as they are the vertexes of the attachment line. Since they could not captured by both cameras, the commissure points were extrapolated along the direction lines of free edge markers.

relaxed valve

156 (frame number)

164

166

167

168

172

176

180

182

186

190

192

200

202

208

210

216

222

230

236

237

239

242

250

Figure 4.10 Reconstruction 3-D configures of free edge

Figure 4.11 shows the displacements of Nodulus Arantii in world coordinate axes. Relatively large displacement from closed position to opened position was observed in the X direction. The displacement was larger during the opening stage and then stabilized during the closing stage. Displacements along the Z direction were mostly positive, but at the end of the valve closing period the free edge was at a lowest Z position due to the turbulent flow.

Figure 4.11 3-D displacements of the point on nodulus arantii

Figure 4.12 shows the X-direction displacement of different markers along the free edge. Markers 2 and 21 were near the commissures and hence their displacements were relatively small. Markers 7, 12, and 17 corresponded to 1/4, 1/2, and 3/4 length on the free edge. Marker 12 was at Nodulus Arantii which had the largest displacement. The displacement pattern of marker 7 was similar to marker 12. However, the displacement of marker 17 was much stable comparatively. It showed that leaflet motion is not symmetrical; the movement of the free edge under the pulsatile flow was more dominant at the center and left side.

Figure 4.12 Displacements of different markers on free edge along x-direction

Chapter 5 Numerical Methodology

5.1 Finite Element Method

When the aortic valve functions during the cardiac cycles, relatively large translation and rotation of the leaflet takes place. The leaflet will come into contact with each other when the valve is in close position. In the current study, to simulate the cyclic dynamic process of the aortic valve, a finite element software (ANSYS 8.1) was used. The ANSYS provides an interface to the LS-DYNA explicit dynamics finite element program. The explicit method-of-solution used by the LS-DYNA provides fast solutions for large deformation dynamics and complex contact problems. Hence, it is suitable for the dynamic simulation of the aortic valve (Thornton MA et al., 1997). The model was first built in the ANSYS interface and later solved by the LS-DYNA solver. The result can be viewed using the standard ANSYS post-processing tools. In the following sections, the background of the finite element method and the steps for solving the dynamic problems are introduced.

5.1.1 Finite Element Background

In the last thirty years, numerical modeling and simulation have become increasingly important in the field of bioengineering (www.csep1.phy.ornl.gov/CSEP/BF/BF.html). Finite element method (FEM) is one of the important simulation techniques because it is capable of modeling problems with great complexity and geometry. It has since, become a powerful tool in finding the solution of critical problems in heat transfer, fluid mechanics, and mechanical systems. Also, the availability of fast and inexpensive computers allows more and more engineering problems to be solved using FEM.

The FEM was introduced by Richard Courant to solve the torsion stress on a cylinder.

Courant's contribution drew on the earlier works of the partial differential equations (PDE) developed by Rayleigh, Ritz and Galerkin. The increase of the FEM usage began in the mid 1950s (for airframe and structural analysis). In the 1960s the FEM was used mostly in the civil engineering (www.wikipedia.org). The first book on finite element by Zienkiewicz and Chung was published in 1967. In the late 1960s and early 1970s, the finite element analysis was extended to solve non-linear problems and problems with large deformations.

In the finite element method, the solution domain is discretized into a number of uniform or non-uniform finite elements that are connected via nodes (www.csep1.phy.ornl.gov/CSEP/BF/NODE8.html). Since the elements can be different in shapes and their connection, the FEM can be used to solve problems with complex geometries. The changes of the dependent variable within the whole solution region are approximated within each element by an interpolation function. The interpolation function is defined relative to the values of the variables at the nodes associated with each element. The nodes variables were called degrees of freedom (DOF). Thus, the problem of solving for infinite continuous DOFs was simplified to solving for finite discretized DOFs. After the values of these variables have been obtained, the approximated solutions at each element can be calculated through the interpolation functions. Accordingly, the approximation result in the whole solution domain can be obtained (Wang XC et al., 1997).

Finite element analysis (FEA) is the application of the FEM to the analysis of static or dynamic physical objects and systems (www.wikipedia.org). A common use of FEA is for the determination of stresses and displacements in mechanical objects and systems. Besides that, it is also routinely used in the analysis of many other types of problems, including those in heat transfer and fluid mechanics.

In the current study, the FEM was used to simulate the dynamics and mechanical properties of the BHVs. The steps for dynamic simulations are as follows :

5.1.2 Steps of Solving Dynamic Problem

The basal equations of 3-D elastic dynamics are listed below.

Equilibrium equation:

$$\sigma_{ij,j} + f_i = \rho u_{i,tt} + \mu u_{i,t} \quad (5.1)$$

Geometry equation:

$$\varepsilon_{ij} = \frac{1}{2}(u_{i,j} + u_{j,i}) \quad (5.2)$$

Constitutive equation:

$$\sigma_{ij} = D_{ijkl} \varepsilon_{kl} \quad (5.3)$$

Boundary conditions:

$$\left. \begin{aligned} u_i &= \bar{u}_i \text{ (at } S_u \text{)} \\ \sigma_{ij} n_j &= \bar{T}_i \text{ (at } S_\sigma \text{)} \end{aligned} \right\} \quad (5.4)$$

Initial conditions:

$$\left. \begin{aligned} u_i(x, y, z, 0) &= u_i(x, y, z) \\ u_{i,t}(x, y, z, 0) &= u_{i,t}(x, y, z) \end{aligned} \right\} \quad (5.5)$$

In equation (5.1), ρ is the mass density; μ is the damp coefficient; $\rho u_{i,tt}$ and $\mu u_{i,t}$ refer to inertial force and damping force respectively. The inertial force and the damping force are the most important characteristics that differentiate the dynamic analysis from static analysis.

The steps of finite element dynamic analysis are as follows:

1. Discretizing the whole solution domain into small finite elements.
2. Creating the interpolation functions.

$$\mathbf{u} = \mathbf{N}\mathbf{a}^e \quad (5.6)$$

where $\mathbf{u}$ is the displacement field within an element, $\mathbf{N}$ is the shape function matrix, and $\mathbf{a}^e$ is the element displacement vector.

3. Forming the governing equation of the whole system

The governing equation can be derived from the Galerkin's equation of the equilibrium and the boundary condition. It is also called the motion equation.

$$\mathbf{M}\ddot{\mathbf{a}}(t) + \mathbf{C}\dot{\mathbf{a}}(t) + \mathbf{K}\mathbf{a}(t) = \mathbf{Q}(t) \quad (5.7)$$

where $\mathbf{M}$ is the mass matrix, $\mathbf{C}$ is the damping matrix, $\mathbf{K}$ is the stiffness matrix, and $\mathbf{Q}$ is the external dynamic loading force vector.

4. Solving the equations of motion

In ANSYS/LS-DYNA, a direct explicit integration method (central difference method) was used to solve the motion equations. The semi-discrete equations of motion used in LS-DYNA (Hallquist JO, 1998) at time step n are:

$$\mathbf{M}\ddot{\mathbf{a}}^n = \mathbf{P}_{\text{ext}}^n - \mathbf{P}_{\text{int}}^n \quad (5.8)$$

where $\mathbf{M}$ is the diagonal mass matrix, $\mathbf{P}_{\text{ext}}^n$ accounts for external and body force loads, and $\mathbf{P}_{\text{int}}^n$ is the internal force vector, which includes stress divergence vector, hourglass resistance, and contact force. The hourglass resistance is non-physical zero-energy modes of deformation. It occurs in under-integrated solid and shell elements.

The central difference time integration was used to advance to time t^{n+1} :

$$\ddot{\mathbf{a}}^n = \mathbf{M}^{-1}(\mathbf{P}_{\text{ext}}^n - \mathbf{P}_{\text{int}}^n) \quad (5.9)$$

$$\dot{\mathbf{a}}^{n+1/2} = \dot{\mathbf{a}}^{n-1/2} + \ddot{\mathbf{a}}^n \Delta t^n \quad (5.10)$$

$$\mathbf{a}^{n+1} = \mathbf{a}^n + \dot{\mathbf{a}}^{n+1/2} \Delta t^{n+1/2} \quad (5.11)$$

where

$$\Delta t^{n+1/2} = \frac{(\Delta t^n + \Delta t^{n+1})}{2} \quad (5.12)$$

$\dot{\mathbf{a}}$ and $\mathbf{a}$ are the global nodal velocity and displacement vectors, respectively. The geometry is then updated by adding the displacement increments to the initial geometry:

$$\mathbf{x}^{n+1} = \mathbf{x}^0 + \mathbf{a}^{n+1} \quad (5.13)$$

5. Calculating for the strain and stress

After the global nodal displacement vector has been obtained using equation (5.2) and (5.3), the strain and stress can be calculated (Wang XC et al., 1997).

5.1.3 Stability of Central Difference Method

The solution is achieved by splitting up the analysis into a large number of small time steps. The time steps are of a size less than the time taken for a sound wave to travel across the smallest element in the analysis (Thornton MA et al., 1997). ANSYS/LS-DYNA checks all elements when calculating the required time step (ANSYS/LS-DYNA User's Guide). The time step has to be small enough to avoid introducing numerical instability into the solution and large enough to allow the

solution to proceed at an acceptable speed. A scale factor of 0.9 was used to reduce the time step:

$$\Delta t = 0.9 \frac{l}{c} \quad (5.14)$$

The characteristic length, l , and the wave propagation velocity, c , are dependent on element type.

For shell elements:

$$l = \frac{(1 + \beta)A}{\max(l_1, l_2, l_3, (1 - \beta)l_4)} \quad (5.15)$$

where $\beta = 0$ for quadrilateral and 1 for triangular shell elements, A is the area, and $l_i (i=1,2,3,4)$ is the length of the sides that define the elements.

$$c = \sqrt{\frac{E}{\rho(1 - \nu^2)}} \quad (5.16)$$

where E is the Young's modulus, ρ is the density, and ν is the Poisson's ratio.

5.2 Modeling of Aortic Valve Leaflet Geometry

To carry out the FEA of the bioprosthetic aortic valve, the geometry of the aortic valve was modeled first, and then transferred into the finite element software for analysis.

In this study, the geometry of the leaflet model is similar to the bioprosthetic valve leaflet (experimental), created using the ANSYS pre-processor. The 3-D locations of the markers on the leaflet in the relax state were used to create the shape (see Figure

5.1). It was difficult to capture all the markers on the concave shape of the leaflet through both cameras in this experiment. The markers that could be seen in both cameras include the markers along the free edge, the markers on the belly area, and the ones along attachment line.

Figure 5.1 Aortic valve geometry

- 1. leaflet commissures; 2. free edge(red line); 3. belly area;
- 4. Nodulus Arantii region; 5. attachment line(blue line)

In order to create the initial shape of the leaflet based on the limited 3-D reconstruction markers, additional information was added. The first addition was some points on attachment line, which were blocked in the paired images of relaxed leaflet. These markers appeared in both cameras during the leaflet opening stage. Attachment line of the valve leaflet was fixed on the stent structure, which was assumed to be immovable. So, the attachment line markers collected from the leaflet opening images also could be used in the initial leaflet geometry. The second addition was the two commissure points. Since the two points were not reconstructed directly, they were extrapolated from the markers on free edge as described in Chapter 4. The third additional information came from the typical characteristic of the aortic valve. Figure 5.2 shows the schematic representation of aortic valve dimension based on Swanson and Clark (Thubrikar, 1990). All the design parameters were the functions of

the base of aortic valve (R_b). The commissure height (H_s) and coaptation height (C_c) were then calculated. H_s/R_b ranges from 0.22~0.7, and C_c/R_b ranges from 0.1~0.34 (Thubrikar, 1990; Beck A et al., 2001; Hart JD, 2002; Ramakrishna G et al., 2002; Ranga A et al., 2004). For the tested aortic valve, the radius of the base was 10mm, H_s was 6mm and C_c was 3mm.

Figure 5.2 Design parameters of valve leaflet

After the H_s and C_c had been obtained, the points below the commissures (M) and the Nodulus Arantii point (N) were approximated. Points M' and N are shown in Figure 5.2. The commissures and point M were then connected respectively. The curve that runs through M and the attachment line around the belly area was approximated with a spline arc. The curve ran from Nodulus Arantii point to the base of the attachment line were created through the similar process. The free edge was formed by a spline-fit curve through all the reconstructed free edge markers. The leaflet area was then constructed using these curves (see Figure 5.3).

Figure 5.3 Geometry of tested BHV leaflet

5.3 Finite Element Analysis of Aortic Valve

5.3.1 Preprocessing

Preprocessing is the first step of the whole simulation. At this stage, the properties of the numerical model such as element type, material properties and real constants were defined. Then, the imported leaflet geometry was divided into several smaller areas for the purpose of producing a good quality mesh. Following, the model was meshed and ensuring that no break occurs at the boundaries between the adjacent areas. As the leaflets coaptate with each other as the valve closed, contact elements were defined. Following that, the loading and the boundary conditions were assigned to the model. These are basic preprocessing requirement to be taken.

The details are described as following.

Element Type and Real Constant

There were several element types that have been used in the literatures. Membrane shell was used in the earlier years (Hamid MS et al., 1986; Chandran KB et al., 1991). This element type is a tension only element and ignores the bending. This is not suitable to represent the leaflet motion since the bending has been proven to be responsible for the failure of the BHVs. Another element type used was brick element (Krucinki S et al., 1993; Hart JD et al., 2002). Since the leaflet is a thin structure, the brick element needs to be very small for successful computation. Furthermore, in order to represent the thickness variation in the leaflet, several layers of brick element will be required. Thus in dynamic simulation where the time step would be relatively small, this can greatly increasing the computational time. Thin shell element with both bending and membrane capabilities had been used widely to analyze the aortic valve. This element type was also used in current study. By using this element, the time step would be independent from the thickness of the shell. Besides that, there are several integration points through the thickness of the shell, from which the variation in thickness could be modelled easily.

In the ANSYS/LS-DYNA, there is only one type of shell element available, i.e. Shell163. However, several formulations were available for selection, such as triangular shell formulations, membrane element formulations, and general shell formulations. As the leaflet was subjected to large deformations including bending, the membrane element formulations were not appropriate. Generally, it is not recommended to use the triangular shell formulations for the entire numerical model due to accuracy reason. Therefore, general shell formulations were used. The Belytschko-Tsay element formulation with one-point quadrature is the default formulation used in LS-DYNA since it is the fastest solution formulation. However, it does not treat deformed configurations accurately. The hourglass effect often occurs in the computation with Belytschko-Tsay one-point integrated formulation. Thus, fully-integrated Belytschko-Tsay shell element formulation was chosen. This

formulation is based on a combined co-rotational and velocity-strain formulation. The efficiency of the element is obtained from the mathematical simplification that results from two kinematical assumptions (co-rotational and velocity-strain). The co-rotational portion of the formulation avoids the complexities of nonlinear mechanics by embedding a coordinate system in the element. The choice of velocity strain in the formulation facilitates the constitutive evaluation, since the conjugate stress is the well-known Cauchy stress. The hourglass effect was avoided with the full integrated formulation. Three integration points were chosen through the thickness. The limitation of formulation was that it did not pass the patch test. Thus it might not converge although the number of the element was increased.

Since the 3-D reconstructed coordinates of the 21 markers on the free edge during the cardiac cycle would be loaded as the dynamic boundary conditions of the FE model, the markers were set as nodes. The mesh size of the whole leaflet was limited by these nodes. The geometry of the BHV was divided into several small parts to enable map-meshing.

Two models with total element number of 916 and 2740 respectively were tested. In the latter model, the middle nodes were interpolated among the markers on the free edge. Then the model was map-meshed based on these nodes. The displacements of these added nodes during the cardiac cycle were also interpolated using cubic spline functions. A smoother deformation was obtained in the finer mesh. However, excessively long computation time was needed. The 916 elements size was found to be time-efficient, and therefore it was chosen for the current simulation study.

The natural aortic valve leaflet has variable thickness. However, the information on detailed thickness distribution is lacking especially for pericardium. In the current numerical simulation, the leaflet was assumed to have uniform-thickness of 0.5mm, which is an average thickness of leaflet tissue (Beck A et al., 2001).

Material Properties:

The material properties in ANSYS are based on the classification of the material's mechanical behavior, e.g. isotropic, orthotropic, linear, nonlinear, etc. The tested BHV aortic valve was a porcine valve where its material property is both anisotropic and nonlinear elastic.

a) Young's modulus

The BHV tissue, whether porcine aortic valve or pericardium tissue, are made from nonlinear materials. For ease in modeling, the stress-strain curves of these tissues are regarded as bi-linear. The segment with lower modulus of elasticity was defined as pre-transition phase, and the segment with higher modulus was defined as post-transition. The pre-transition modulus was found to range from the lowest at 11kPa (human leaflet) to the highest at 245kPa (canine leaflets) , Thubrikar MS (1990). And the post-transition modulus ranged from 1MPa to 6MPa. Thus in the post-transition phase there is a steep change in the modulus after the pre-transition stage. Several non-linear materials in LS-DYNA material library were also examined. The commonly used hyperelastic material type was the Mooney-Rivlin rubber model. This model allows a least-square-fit of the stress-strain curve input. However, the fitted curve of Mooney-Rivlin rubber only has a gentle slope; making it unsuitable to model the natural tissue steep stress-strain curve. Besides that, Mooney-Rivlin model was not suitable for a large bending simulation since it was designed for rubber which bears tensile loading. Another material model was a soft tissue model: material type 91 in LS/DYNA (Weiss JA et al.,1996; Quapp KM et al., 1998). The model is basically an isotropic Mooney-Rivlin matrix reinforced by fibers. Stress-strain curves (Li J et al., 2001) were used to estimate the parameters for this soft tissue model. Because this material type could only be used with one-point integrated Belytschko-Tsay formulation, the hourglass effect could not be avoided.

Generally, low pre-transition modulus was thought to assist the leaflet opening, and high post-transition could prevent excessive deformation. But it was found that the numerical model with the low pre-transition modulus experiences too much deformation that the computation could not converge. And the material with high post-transition modulus makes the leaflet difficult to open. Since the nonlinear material types were not good enough for the current simulation, the linear material properties were then used. In the current study the Young's modulus was assumed to be 1MPa as the average value to ensure a complete calculation within a whole cardiac cycle. The similar assumption of 1MPa Young's modulus was also used by Ramakrishna G et al. (2002), while Candra J et al. (2002) used 900kPa. Indeed, it is difficult to accurately represent the true material properties of the valve tissue. This simplified isotropic-linear-material assumption is a limitation in this study.

The valve stent which supports the valve tissue has very little movement during the cardiac cycle, thus it was reasonable to assume to be rigid. Since the leaflet motion is almost independent from the aortic wall because of the stent, the FE models do not include the aortic wall.

b) Density

The mass matrix is an important part in the motion equation. And the density of the valve leaflet is therefore essential. In open literature, the density of the heart tissue were seldom reported. Kunzelman KS (1998) reported the density of mitral valve to be 1.04g/cm³. On another analysis Ramakrishna G (2002) used 1.1g/mL (1mL = 1cm³) as the density of the leaflets and 2.0g/mL as density of sinus, but no data source was mentioned. Since the reported values are vague, in order to obtain a more reliable values, an experiment was carried out to determine an average value of the density of the tissues.

In this experiment, three fresh porcine hearts from wet market were used. The aortic

leaflets, mitral leaflet, and aortic wall were carefully excised. The sample size for mitral valves was four (3 anterior and 1 posterior leaflet), for aortic valve leaflets was 9 pieces, and for aortic roots was 3. A piece of pericardium was also excised for this measurement.

The mass of each type of tissues were measured by a digital quantity measuring device, Precisa balance (Precisa Instruments Ltd, Switzerland). The range of measurement is 200g, with an accuracy of 0.0001g.

Because the leaflet tissues are very soft, thin and irregular, it was hard to measure the volumes directly. The water displacement method was chosen for volume measurement. Since the tissues were different in shape and volume, three different measuring cylinders were chosen for measurement: 3ml, accurate to 0.1ml; 10 ml, accurate to 0.2ml; 60ml, accurate to 1ml (syringe); 300ml, accurate to 2ml.

The measured data of the three types of tissue are listed in Table 5.1. Since the mass measuring device is sensitive and accurate, the main error arises from the volume measurement. Maximum relative error is less than 6%.

	Mass (g)	Volume (mL)	Density (g/mL)	Maximum Relative error
Mitral Valve	1.7684 ± 0.0001	1.7 ± 0.1	1.04 ± 0.06	6%
Aortic Valve	1.0957 ± 0.0001	1.05 ± 0.01	1.04 ± 0.01	0.7%
Aortic root (total)	11.4790 ± 0.0001	11.0 ± 0.2	1.04 ± 0.02	2%
Pericardium	2.0944 ± 0.0001	2.0 ± 0.02	1.05 ± 0.01	1%

Table 5.1 Measurement for density

Comparing with density value reported by Kunzelman, the measured results are

logical and within the tolerance range.

In conclusion, from the experiment, it was found that the densities of the three tissues (mitral leaflet, aortic leaflet and aortic wall) are very similar, even though they are different in shape, thickness and tissue structure. Density of 1.04g/mL was then used in this numerical study.

c) Poisson's ratio

To represent the incompressible properties of the tissue, a Poisson's ratio of 0.45 was used (similar to previous studies).

Contact condition

When the aortic valve closes, the three leaflets will come into contact due to the diastolic pressure difference between the aortic and ventricular pressures. This coaptation will prevent reverse flow from the ascending aorta back to the ventricle. In numerical simulation, if the contact condition is not defined, then the leaflets would pass through each other. In this study only one leaflet of the valve was created. And a fictitious surface was modeled in the region where the leaflet is to coapt (see Figure 5.4). The reconstructed 3-D coordinate markers along free edge at the diastole were used to create the contact target surface. Spline curve was created through these marker points and extended a little on both sides. The spline curve was then extruded along Z axis to form the fictitious surface. The coaptation of two leaflets was replaced by the leaflet contacts with the surface. Surface-to-surface contact type was chosen for this simulation.

There is no information or investigation relating to friction between the coapting leaflets. Since the aortic valve moves in a liquid environment (blood) and the surface of the leaflet is extremely slippery, a frictionless sliding contact was considered in the

modelling. Previous researchers who applied frictionless contact in similar modeling include Patterson EA et al., 1996; Grande KJ et al., 1998; and Chew GG et al., 1999.

Figure 5.4 Meshed leaflet and its coaptation surface

Loading

Though the leaflets loadings vary both spatially and temporally due to the blood flow, there has been little information regarding the spatial variation of pressures on the leaflets.

The loading conditions were simulated with the experimental time-varying physiological pressures applied to the leaflet surface. Aortic pressure and ventricular pressure were recorded whilst the motion of leaflet was being captured. The sample

time for pressure recording was 0.0032sec. In order to synchronize the pressure to the images captured, the pressure curves were interpolated using cubic spline functions. The difference between the aortic pressure and the ventricular pressure was applied perpendicular to leaflet surface on the ventricle side (see Figure 5.5).

Figure 5.5 Experimental left ventricular and aortic pressures and their difference

The period from the valve opening to closing is the actual ejection time, and the period in which the pressure difference is larger than zero is the catheter ejection time (Garcia D et al., 2005). The catheter ejection time is always shorter than the actual ejection time. In the numerical simulation, when the pressure difference was lower than zero, leaflet would close. During the middle stage of systolic phase, vortex effect is an important factor that influences the valve motion; it maintains the valve opening configuration (Korakianitis T et al., 2005). Since the vortex effect could not be present in the catheter pressures, in order to maintain the aortic valve opening in the numerical simulation the ventricular pressure was multiplied by 1.13 (so that the ventricular pressure matches with the aortic pressure at dicrotic notch as reference). Arcidiacono G enlarged the pressure difference by introducing a “reduced aortic

pressure” equal to 75% of the real aortic pressure (2005). But the reduced aortic pressure would cause the pressure difference in the diastole to be much smaller. The enlarged ventricular pressure (see Figure 5.6) used in current research would help the simulation in systolic phase with relatively small effect during the diastolic phase.

Figure 5.6 Original and modified pressure difference

As the initial geometry in this study was in a relaxed configuration, an initial pressure-ramp was applied starting from end diastolic phase. To keep an effective computing time, the time for adding ramp pressure was relatively short. However, if the period of the pressure-ramp is too short, the unnecessary large response may affect the later cyclic simulation. In the study the pressure difference was applied from 0 to the end of diastole in 100 msec (refer to Figure 5.6). Then, the period of cardiac cycle began.

Boundary conditions

Majority of BHVs are designed with a valve stent that provides structural support for the leaflets. This design configuration allows the BHVs attachment process to the aortic wall more convenient. The valve stent, although made from flexible material, still impose rigidity to the adjacent natural aortic root and wall. In the numerical model, the stiffness condition of the valve stent was modeled by fixing all the nodes along the attachment line so that they could neither rotate nor translate.

In the current research, the moving leaflet under pressure difference (only) was simulated first. In this simulation, the coaptation surface was assumed to be rigid and wholly fixed. Next, the motion of the leaflet under both pressure difference loading and dynamic boundary conditions (along free edge) was simulated. When the leaflet closes, it will come in contact with the coaptation surface. Since part of the free edge would cross the coaptation surface while leaflet was moving, the coaptation surface was assumed to be of the same tissue as the leaflet and bear the same pressure difference. The fixed attachment line of the leaflet and the boundaries around the coaptation surface were shown in Figure 5.7 with red pressure contour on the surface.

Figure 5.7 Boundary conditions and pressure load

Figure 5.7 also shows the dynamic boundary conditions applied on the nodes along free edge. The displacement curves on the three world coordinate axes were calculated and applied on the nodes. The displacements of each node at each time-step were calculated by subtracting the initial relaxed location of each node from the

current time-step 3-D location. The reconstructed 3-D coordinates covered only 95 frames (380msec) as the leaflet experienced similar configuration during the rest of diastolic phase.

The aortic valve moved under the pulsatile pressure. The leaflet is “pushed” by the fluid flowing from the base of attachment to the free edge. The vibration of free edge obtained from experiment was the result of the flow. In the current study, the motion of the free edge was applied as the boundary condition in order to improve the accuracy of the numerical simulation. If the displacements were applied in the normal time sequence, the motion in the leaflet would start from the free-edge to the base of attachment. This would be a wrong waving direction. Thus the dynamic boundary conditions were modified by reversing the time sequence, i.e. the displacement on the first time was placed on the last time (see Figure 5.8). During the computation, the waves were also produced from the free-edge to the attachment line. But in the post-processing the results were sequenced from the last time to first time (checked in a reversed time sequence). Hence, the leaflet waved from the attachment to the free edge (the vibration sequences of the free edge were the same as the natural sequence). Thus after the reverse process, the natural waving function was obtained in the numerical simulation. As the dynamic boundary conditions, which were applied on free edge node, were in a reversed time sequence, the pressure load on leaflet surface were also applied in a reversed time sequence (Figure 5.9).

Figure 5.8 3-D dynamic boundary conditions on one node

Figure 5.9 Pressure difference load in a reversed time sequence

5.3.2 Solution

After the complete preparation of the finite element model, the models were solved

using the LS-DYNA solver. The total solution time was 956msec (100msec ramp load period+60s/70beat).

5.3.3 Post-processing

The LS-DYNA results could be analyzed in the standard ANSYS post-processing. This study focused on the deformation and the stress distribution of the leaflet over time and space. The results are presented in details in the next chapter.

Chapter 6 Numerical Simulation Results

The main contribution of this dissertation was the application of dynamic boundary conditions for FEA of a bioprosthetic valve. As a result, a novel experimental technique was developed to capture the profile of the valve leaflet under pulsatile conditions. Comparison was made with conventional approach by considering only the time-varying uniform pressure load on the leaflet. In this chapter, the results of the two different models (the deformation, the stress distribution, and their bending stress) are shown and compared.

Model 1 refers to the leaflet model with only pressure load applied.

Model 2 refers to the leaflet model with both the pressure load and the dynamic boundary conditions applied on the free edge.

6.1 Deformation

Figure 6.1 compares a series of leaflet deformation for Model 1 and Model 2. And Figure 6.2 shows the detail deformations of Model 2 and original valve leaflet movement obtained in experiment. The frame number of the experimental results which corresponds to the time instance of simulation results is indicated beside each image.

From Figure 6.1, it can be seen that at each same instance, the two simulation results show significant difference. Generally, the leaflet in Model 1 opens a little earlier and closes much earlier than Model 2. From time frame 210, Model 1 leaflet moves backward to close and there is a small opening later at frame 216 while in Model 2, the valve remains fully open till time frame 216.

Initial shape

Left: Model 1

Right: Model 2

157

163

164

167

208

210

216

239

250

Figure 6.1 Comparison of leaflet deformation during a cardiac cycle for Model 1 and Model 2

relaxed valve

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Figure 6.2 Deformation of leaflet during one cardiac cycle (Model 2)

Figure 6.1 also shows that the aortic valve is fully closed at the beginning of the systolic period due to the relatively high pressure difference. Then, the valve relaxes as the pressure difference is lowered (frame 163), and it opens when the ventricular pressure is greater than the aortic pressure (frame 164). A very short time frame was needed for the area of the leaflet orifice to reach maximum opening (frame 167). The leaflet then maintains similar configurations for about 172ms till frame 210. Thereafter, the leaflets move back slowly until they are almost closed (from frame 210 to 239). The leaflet is shown to be in fully closed position from the time the dicrotic notch appears until the end of this cardiac cycle (frame 250).

Figure 6.2 shows the simulation of Model 2 with the corresponding experimental images. Under the influence of the imposed dynamic boundary conditions, the leaflet moves in more complicated configurations. The valve leaflet closes tightly at the beginning of systolic period, and then relaxes and is seen to start to open from frame 164. The leaflet starts the opening process from the belly area to the free edge (frame 164-170). After about 28ms (frame 170), the leaflet opens to its maximum configuration. Then, the leaflet draws back a little and sways in the fluid flow (frame 170-226). That part of leaflet that swings back and forth is mainly from the middle section of the leaflet to the free edge, with the free edge having the largest sway. At the later period of the systole, the valve leaflet stops swaying and moves back slowly as fluid flux lowers down (from frame 226 to 234). The leaflet keeps a similar configuration in this stage (having a little movement). A quick closing of the leaflet starts from frame 234 to frame 238. Though the free edge of the leaflet is dragged behind, the middle area of the leaflet closes tightly for the first time. The valve is then fully closed after the time when the dicrotic notch appears. This valve closure position remains throughout the remaining diastole period.

6.2 Stress and Bending Moment Distribution

It was reported that the main damage of BHVs occurred largely due to mechanical

reasons (Vesely I et al., 2001). Cuspal rupture is related to flexural fatigue and the presence of compressive stresses on the inner surface of the bent tissue. Therefore, the distribution of the leaflet stress and bending moment are important.

The effect of externally applied forces on a body is the generation of internal forces within the body to resist or balance the effect of the external forces (www.iago.stfx.ca). The intensity of the internal forces or force per unit area is called stress, which represents the resistance of materials to deformation. At a given point on a surface, a stress vector depends on the orientation of the surface. The stress vector on three mutually perpendicular planes at a point specifies the state of stress at that point in a continuum. The components of these stress vectors form a tensor, the stress tensor. Mathematically, the stress tensor is a second-order Cartesian tensor with nine stress components. Three stress components that are perpendicular to the planes are called normal stresses. Those components acting tangent to these planes are called shear stresses (www.documents.wolfram.com).

The normal stresses and the shear stresses vary smoothly with respect to the rotation angle, in accordance with the coordinate transformation equations. There exist a couple of particular angles where the stresses take on special values. There also exists an angle where the shear stress becomes zero. The angle defines the principal directions where the only stresses are normal stresses, and the first and third principal stresses are the maximum and minimum normal stresses respectively at an arbitrary point (www.efunda.com). The second principal stress value is between the first and third principal stresses, and it is not as significant in terms of engineering viewpoint. The first principal stress is marked as S_1 , and the third principal stress marked as S_3 . From the distribution plots of the first principal stress and the third principal stress, the locations with the extreme tensile and compressive stresses can be determined.

In this study, shell element was used to model the thin valve structure. Because the leaflet has a large deformation during the cardiac cycle, the stresses were expected to

vary a lot through the thickness. There are some integration points through the thickness that can be used to model these large-variation stresses. Originally, the through-the-thickness integration points were used in the integration to obtain the element stiffness matrix. Typically, 2 integration points are required through the thickness for elastic behavior, while 3 or more integration points for plastic behavior. In the current simulation, two integration points are sufficient for linear elastic material. The through-the-thickness integration was used in the calculation of the bending moments and the in-plane forces.

6.2.1 Stress Distribution

First the contour plots of the first principal stress in the leaflet, which is also called the maximum principal stress (S_1), are illustrated in Figure 6.3. This, also called membrane stress, was obtained from the three integration points (along the thickness of the shell element). Similar illustrations were also presented by Chew et al. at 1999 and Ramakrishna et al. at 2002. Figure 6.3 shows the membrane stress distribution of the Model 1 during the whole cardiac cycle, and Figure 6.4 shows the stress distribution of Model 2.

In figure 6.3, the membrane stresses during systole period are comparatively smaller than those in diastole period. During this period, the high stress region mainly along the free edge and in the belly area. In the diastole period, when the valve closes tightly due to large pressure, the membrane stresses increase relatively a lot in the whole leaflet. The highest stresses are found around the free edge.

In Figure 6.4 high membrane stresses occur at commissure region and along the free edge during the whole cardiac cycle. During the period that the leaflet opens, stress values are less than 300KPa. When the leaflet closes, the membrane stresses at the commissure increase to 540KPa, and the stresses reach 1MPa during the diastole period.

Figure 6.3 S1 (GPa) in the mid-surface of the leaflets through the cardiac cycle (Model 1)

The above figures show the maximum principal stress $S1$ at the mid-surface of the shell element. They may not be sufficient to illustrate the entire stress state for the valve because of bending; the stresses vary a lot along the thickness of the leaflet.

The failure of the leaflet is also related to the compressive stress, which can be represented by the third principal stress ($S3$). Two principal stresses in the inflow, middle, and outflow surface of one of the opened leaflet configurations are shown in Figure 6.5. For comparison, all the six plots use a uniform stress contour range. The stress range is from the minimal $S3$ value to the maximal $S1$ value.

Figure 6.5 $S1$ and $S3$ (GPa) distributions on different surface level in the leaflet

It is observed that the color contour plots of the stresses in the middle surface are quite evenly distributed. The absolute stress values in the other two surfaces are relatively higher. The largest $S1$ corresponds to the outflow surfaces while lowest $S3$ corresponds to the inflow surface. In the outflow surface, the largest $S1$ are located at the upper part of the attachment line (near the commissure region, on the left side of the leaflet) and lower part of attachment line (at the belly area on the right side). The same regions in the inflow surface show the lowest $S3$. The upper part of the attachment line near the commissure region on the right side of the leaflet also has a

relatively low S3 value in the inflow surface. The higher absolute stress at the inflow or outflow surface than the mid-plane surface is resulted from bending moment. It was noticed that stress values along free edge are similar through all the three layers, indicating that the main stress type along the free edge is the membrane stress.

6.2.2 Bending Moment Distribution

Bending is one of the most important causes for the failure of the bioprosthetic aortic valve. Though S1 and S3 can illustrate the maximum tensile and compressive stress distribution, which can provide some information about bending, they can not describe the bending status, such as magnitude, clearly. Therefore, the effective bending moment per unit length was introduced, which can be defined as

$$M = \sqrt{M_x^2 + M_y^2 - M_x M_y + 3M_{xy}^2}$$

where $M_x = (\sigma_x^{\text{top}} - \sigma_x^{\text{bottom}})h^2 / 12$, $M_y = (\sigma_y^{\text{top}} - \sigma_y^{\text{bottom}})h^2 / 12$ and

$M_{xy} = (\tau_{xy}^{\text{top}} - \tau_{xy}^{\text{bottom}})h^2 / 12$, and h is the thickness.

M_x , M_y and M_{xy} can be obtained from the ANSYS postprocessor using the stresses between inflow and outflow surfaces. 'x' and 'y' refers to the element in-plane coordinate system. The contours of the bending moment are shown in Figure 6.6 (Model 1) and Figure 6.7 (Model 2). Similar bending moment contour was also illustrated by Li at 2001. Though this contour is from the absolute value, the bending direction can be determined from the deformation.

Figure 6.6 Distribution of bending moment per unit length of the leaflet. The values on the contours are given in kN mm mm^{-1} (Model 1)

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170

164

172

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Figure 6.7 Distribution of bending moment per unit length of the leaflet. The values on the contours are given in kN mm mm^{-1} (Model 2)

Figure 6.6 shows that the bending moment in Model 1 is relatively small (lower than $7 \times 10^{-3} \text{ N mm mm}^{-1}$) when leaflet is still in closing position. The bending moment increases as the leaflet opens, reaching a largest value of $\sim 14 \times 10^{-3} \text{ N mm mm}^{-1}$, and it decreases as the leaflet began to close. Though the free edge bears the high membrane stress, the bending moments along it are not relatively large. During the systole period, relatively large bending moments take place at the two commissure regions that run through the belly area. The largest bending occurs in this belly area. The central area, where the leaflet curvature is inversed, also bears bending stress, but the values are not as large as the belly area below it. When the leaflets close, the bending occurs along the whole attachment line and at the coaptation areas, but the values are smaller.

In Model 2, the difference of bending values in the leaflet between the systole and the diastole stage is larger than in Model 1. In the diastole period when the leaflet is still close, the bending moment is lower than $4 \times 10^{-3} \text{ N mm mm}^{-1}$. The bending moment increases as the leaflet opens. The largest value is around $20 \times 10^{-3} \text{ N mm mm}^{-1}$. The upper part of attachment line near commissure of the left leaflet side bears large bending during systole period. But the largest value does not always occur on this area. The other three parts that also bear large bending are the commissure region of the right leaflet side, the lower part of attachment line at right leaflet side, and part of belly area. Relatively small bending moments occurs along the free edge. The values of bending decrease as the leaflet began to close. And when leaflet is fully closed, high membrane stress with low bending stress starts to dominate.

6.3 Discussion

Comparing Model 1 with Model 2, it can be concluded that although the same initial geometry, meshes, and pressure load were used, the additional imposed boundary conditions causes the simulation results to be quite different.

6.3.1 Deformation

In Model 1, the leaflet opens and closes only under the time-varying and uniform pressure load. The simulation of leaflet in Model 1 appears to be quite stiff, while the

deformations in Model 2 are more flexible (more similar to the real one) because of the imposed dynamic boundary conditions.

Besides the flexibility of the leaflet, Model 2 shows some results that do not show up in Model 1. Firstly, in Model 1 when the valve begins to open, the leaflets do not open so immediately. Actually, since the area of the leaflet does not bear the same ventricular pressure before the leaflets open, the ventricular pressure applied on the coaptation area should be less or later than on the belly area. That difference causes the leaflet to open first in belly area and then followed by the coaptation area and the free edge. Model 1 cannot simulate this effect. In Model 1, the coaptation area opens first. Moreover, the time for real valve opening is longer than the one simulated in Model 1. At the beginning of systole, the ventricular pressure increases very fast, so the time needed for leaflet Model 1 to open fully is only 16ms. In Model 2, the deformations of the leaflet are limited by both the pressure load and the free edge boundary conditions. The time consumed by the leaflet Model 2 to fully open is about 28ms (from frame 164 to frame 171). In addition, the deformations of Model 2 mimic the configurations of real opening process.

Secondly, during the leaflet opening period, the deformations of Model 1 in Z axis are lower than Model 2 because Model 1 maintains a fully opened leaflet shape. After the aortic valve has been fully opened, the vortex effect will be most important factor to influence the valve motion (Korakianitis T et al., 2005). As the results, the valve leaflets sway in the fluid flow causing a comparatively smaller orifice than the maximum opening configuration. Besides that, the position of coaptation area in Z axis should be higher than in the fully-opened stage. Model 2 gives a similar swaying motion during this period, while Model 1 does not.

Thirdly, when the leaflets are closing in, a relatively faster closing process can be simulated in Model 2, while the leaflets of Model 1 experience a slower closing process. In the real valve, since the valve motion is affected by the vortex, the valve maintains the opening longer and does not close as early as Model 1. The leaflet closes fast near the end of the systole, where the pressure effect instead of vortex effect becomes the dominant factor that influences valve motion. The greatly decreased ventricular pressure reverses the direction of the pressure effect, and

induces the fast closing of the leaflets. The leaflets first close in the lower part of the coaptation area, while the free edge is being dragged and formed a large bending across the coaptation area.

The above analysis shows that the deformations arising from the numerical model with dynamic boundary conditions illustrates more realistic simulation.

6.3.2 Stress and Bending Moment Distribution

It is observed that high membrane stresses occur along the free edge in both models due to large deformation. However, high-membrane-stress region (belly region) in Model 1 does not show up Model 2. During the valve opening period, the upper region of Model 1 leaflet reverses its curvature and is driven down under the pressure force. The lower belly area was pressed by both the surface pressure and the inner pushing force from the surface. In Model 2 there is less inner pushing force to the lower belly area, because the free edges wave in a higher position than in Model 1. Thus in Model 2, membrane stresses (in the belly area) are not high.

At the end of systolic period in Model 1, stresses decrease as the leaflet moves back to close. But in Model 2, membrane stresses increase their values as the leaflet moves back. What causes the increase is that when the leaflets are closing fast, the free edge swung back off the closed middle part (coaptation region). Leaflet stretching becomes large in the coaptation area, especially at commissure region and along the free edge.

Though the stress distributions in Model 2 are reasonable, its values may be larger than the real ones (especially the values around the free edge). The real valve functions under the fluid with no restriction on the free edge. But in Model 2, in order to supply the lack of information on the fluid flow, the imposed dynamic boundary conditions are applied on the free edge. Though the displacement data were extracted from the experimental work, they may introduce extra stresses in the simulation.

Since bending is an important factor affecting the deterioration of the aortic valve, it is necessary to understand the bending distributions. In both Model 1 and Model 2, the bending moment is much larger during the valve opening than during the valve

closing. The maximum bending moment in Model 2 ($20 \times 10^{-3} \text{ N mm mm}^{-1}$) is larger than the maximum value in Model 1 ($14 \times 10^{-3} \text{ N mm mm}^{-1}$) due to the larger deformations in Model 2. During the diastolic period, the bending moment is slightly smaller in Model 2 than in Model 1. In Model 2, the bending moments appear asymmetric with relatively complicated distributions. High bending moments mainly occur in four regions: two commissure regions, the lower part of attachment line at right leaflet side, and part of belly area.

Since the failures of BHVs are related with tissue degeneration caused by the mechanical stress, comparing the high stress/moment regions to the failure regions of the corresponding BHVs can help to validate of the simulations in current study.

Calcification is another main reason for the tissue degeneration and failure. Calcium deposits in the leaflet occur almost in all implanted BHVs (usually after 4 years) (Thubrikar MJ. 1990). Degenerative leaflet calcification is prominent at the valve commissures and basal attachment of the leaflet. These two locations also bear the largest bending moment. There are other modes of failure in BHVs besides the degenerative calcification, including tearing and perforation in the leaflets. Ishihara et al (1981) classified the leaflet tears into four patterns (see Figure 6.8). The most frequent tear occurs in the free edge (type I) near the commissure. This tear occurs as a result of collagen breakdown. Linear perforation along the basal regions of the leaflets (type II) is another form of failure. Large oval perforation in the central part of the leaflet (type III) is mostly associated with infection. Pinhole perforations in the middle of the leaflet (type IV) are seen in connection with calcific deposits. Type I tearing is likely to be caused by the stress concentration at the commissure while the large bending running through the belly areas in the open leaflet might lead to the other three tearing types.

Figure 6.8 Four patterns of leaflet tears classified by Ishihara et al

Figure 6.9 Tearing of the Stented valve along the attachment line

Compared with findings in current investigation, the commissures region of the leaflet is indeed a vulnerable region due to high deformation, stress, and bending moment. Though high stresses occur along the free edge, its bending moments are small. In conclusion, all the regions with high bending moments will have a high tendency to rupture of their collagen architecture leading to failure.

Chapter 7 Conclusion and Future Works

7.1 Conclusion

Recent numerical investigations of the aortic valve have advanced to the dynamic simulations involving the moving valve leaflet under the pulsatile flow conditions. However, these simulations were limited because the actual deformation of the valve leaflet (folding and unfolding of the valve leaflet) remains artificial.

In this research, a dynamic numerical simulation was carried out and correlated with the experimental data as a boundary condition. A novel experimental technique was developed to capture the actual valve deformation and these experimental results were applied as the dynamic boundary condition in the numerical simulation. In this approach, a more realistic dynamic deformation and stress/bending of the valve leaflet can be obtained. In summary, the work involves:

- 1) Setting up a pulsatile flow test rig, capable of replicating a “near-physiological” condition of human cardiovascular system. The test section, where the aortic valve was placed, was designed to allow the real-time observation of the motions of the BHV using two charge-coupled device (CCD) cameras through two borescopes.
- 2) Developing an imaging system that enables the two CCD cameras to capture the whole cyclic motion of the BHVs synchronously without the two cameras moving. The cameras’ high-sampling-frequency ensures that thorough details can be recorded throughout the cardiac cycle. A properly chosen shutter speed allows the images captured to have enough light with minimum blur effects. An algorithm was developed to further synchronize the images off-line.

- 3) Designing BHV leaflet markers; small dots were marked on one of the BHV leaflets. These dots were used to trace the leaflet deformation.
- 4) Designing calibration structure. The calibration structure has a comparable size with the tested BHVs. The 3-D locations of the markers on the calibration structure were used as known parameters to reconstruct the 3-D locations of leaflet markers.
- 5) Image processing. The 3-D coordinates were reconstructed using the two synchronous 2-D pictures. Direct linear transformation was used in the reconstruction.
- 6) Creating the finite element model to simulate the tested BHVs using ANSYS/LS-DYNA. The tested valve in a relaxed state, which is assumed to be stress free, was chosen as the initial geometry of the numerical model. The pressure loadings were taken from the pressure curves acquired from the experiment. The pressure curve was then modified to match the catheter ejection time with the actual ejection time. The displacements along the free edge (also obtained from the experiment) were used as the dynamic boundary conditions.

During the cardiac cycle, the aortic valve was deformed due to the fluid loadings on the leaflet surface. The fluid loadings consist of local fluid pressure and fluid shear force. The local fluid pressure data distributed on leaflet surface was difficult to be recorded. Thus, the pressure data used in numerical simulation are the ventricular and aortic pressures which were acquired by the pressure catheter at the upstream and downstream of the valve. The fluid shear force is also difficult to be obtained directly. So, the valve displacement data was used to complement the lack in fluid loading information. Since valve leaflet area will be blocked 'here or there' during the cardiac cycle, only the free edge motion was recorded fully during the cycle. The movement of the free edge was applied as the dynamic boundary condition in the finite element model to give a more accurate simulation result.

The real displacements sequence was due to the leaflet waving from the base of the attachment line to the free-edge. If the boundary conditions are applied at the free edge, the wave movement will be from the free edge to the base of the attachment line. Hence, in order to simulate the real wave movement (which is from the attachment line to the free-edge), both the pressure loading and the dynamic boundary condition was applied with a reversed time axis. The results were then re-reversed in the post-processing stage to represent the wave movement in a correct direction.

- 7) Comparing the results of a model without the dynamic boundary conditions against the proposed model (with dynamic boundary conditions). The comparison shows that the proposed model yield more realistic results.

By applying only the pressure load on the aortic valve model (model 1), at end of the diastolic phase, the leaflet opens initially from the free edge. But the proposed model (model 2, with both pressure loads and dynamic boundary loads) it shows that the leaflet belly area is first pushed up by the ventricular pressure. Then, the lifted-up area push each other at the coaptation area and produce a gap at the center. Finally the free edge is brought along to a valve fully opened position. At the end of the systolic phase, the leaflet starts to close from the bottom line of the coaptation area and then into fully closed position Though the leaflet closing process in model 2 is similar to that in model 1, in model 2, it shows that the leaflet closes more quickly. The quick closing process causes the free edge to fall further behind although the bottom line of the coaptation area has closed. Additionally, during the systolic phase, model 1 cannot show the leaflet swaying motion in the fluid, while model 2 is capable of demonstrating this realistic effect.

The membrane stresses were much larger during the diastolic phase, and relatively low in the opened valve. High stress is distributed along the free edge and at the commissure regions.

The high-stress regions were asymmetric during the cardiac cycle. The regions, where the stress and the bending moment are relatively high, correspond to the frequently failed locations in the BHVs. The quick closing process of the valve is an important phase related to the failure of BHVs because both the bending stress and membrane stress are relatively large at this period.

The main objective of this PhD thesis has been achieved. A novel engineering technique involving capturing of dynamic 3-dimensional data of the valve leaflet in opening and closing has been successful carried out. These data were input as boundary conditions to a FEA software and illustrated more realistic results.

7.2 Limitations and Future Work

The deformations of the numerical model with additional dynamic boundary conditions (model 2) are much similar to the real valve than the model with only the pressure loads considered (model 1). However, there are still limitations in the current model (model 2):

Not *all* boundary conditions were added into the numerical model. In the future work, different positions of borescope can be used in order to capture the more detailed information of the tested valve.

Also, since the displacements were applied along the free edge, where in the real moving valve there is no displacement restriction, the stress values might be higher than they should be.

Next, the leaflet material was assumed to be isotropic, which is actually not sufficient to describe the complex material properties of the valve. Modeling the complexity of

the BHV material is still an ongoing problem. In the future research, the material properties should be determined accurately, so that the anisotropic/nonlinear properties of the valve tissue can be taken into consideration. New code development is also necessary to model the biological soft tissue correctly.

In addition, though the dynamic boundary conditions can help to simulate a more similar valve movement as the real one, they are only the substitute for the missing information of the real fluid loadings. There are two ways to tackle this issue. The first one is developing a coupled fluid-structure simulation model, which allows the structural elements to interact directly with the fluid flow. Secondly, the dynamic deformations that can be captured and reconstructed include not only the free edge but also the other parts of the leaflet. If the deformation of whole leaflet can be reconstructed during the physiological cardiac cycle, even without the pressure load, the strain/stress can be calculated directly.

In addition, the improvements in the 3-D reconstruction technique are also needed. In this study the leaflet markers in each image were still recognized manually because the noises in the images are quite significant; the images were relatively dark and it is difficult to distinguish between the noises and the marker points. In the future, additional image processing methods can be used to remove the noises in the image and automatically recognize the markers on the test object.

Lastly, the tested BHV was a stented one. In the future work, a stentless BHV may be tested but first the aortic root in the physiological mock flow loop should be modified to rubber to accommodate the stentless BHV. Then, the comparison between the stented valve and the stentless one can be carried out.

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