

MEASUREMENTS OF FLOW THROUGH A BILEAFLET
MECHANICAL HEART VALVE IN AN ANATOMICALLY
ACCURATE MODEL OF THE AORTA

by
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Abstract

The objective of this research is to experimentally investigate the flow characteristics past a bileaflet mechanical heart valve (BMHV) in an anatomical model of the aorta. The measurements were made within a mock circulation loop that produced physiological pressure and flow conditions of the aorta. The velocity was measured upstream and downstream of the valve at single points using laser Doppler velocimetry and on planes using planar particle image velocimetry. Viscous and turbulent stresses were evaluated as indicators of potential blood damage. Measurements were first made with a BMHV mounted at the inlet of an axisymmetric channel, which was similar in geometry to channels previously used, and then with the BMHV mounted at the inlet of an anatomical model of the aorta. By comparing these results, the effects of the anatomical shape of the aorta on the flow past the valve were determined. It was found that the level of turbulence past the valve was significantly greater in the axisymmetric model and that the shape of the anatomical aortic sinus, in particular, was effective in reducing turbulence. Additionally, measurements with the valve mounted in three orientations at the inlet of the anatomical aorta showed that the turbulence and the viscous stresses past the valve were lower when the valve was positioned such that its line of symmetry was parallel with the plane of aorta curvature than when it was normal to it. It was further found that flow in the right coronary artery was highest when the valve was positioned with its central orifice aligned with the opening to this artery. The results of this research may be used to assist surgeons in choosing the best implantation orientation of a BMHV.

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Nomenclature

(Units have been indicated in parentheses)

a	Radius of curvature (m)
c_1, c_2	Experimental constants (dimensionless)
D	Outer valve diameter (m or mm)
D_i	Inner valve diameter (m or mm)
d_i	Channel inlet diameter (m or mm)
d_p	Particle diameter (m or μm)
d_{pl}	Distance between PIV measurement plane and calibration plane (m or mm)
De	Dean number (dimensionless)
Ds	Mechanical dose function (dimensionless)
E	Strain rate (s^{-1})
f	Probability density function (dimensionless)
K	Number of bins (dimensionless)
M	Number of cycles (dimensionless)
N_{bk}	Number of measurements in a bin (dimensionless)
P	Pressure (N/m^2 or mmHg)
P_a	Pressure downstream of BMHV/aortic pressure (N/m^2 or mmHg)
P_{bulb}	Bulb pressure (N/m^2)
P_v	Pressure upstream of BMHV/ ventricular pressure (N/m^2 or mmHg)
Q	Volume flow rate (m^3/s or L/min)
q_{xz}, q_{yz}	Planar velocity magnitudes on the x - z and y - z planes (m/s)
q'_{xz}	Planar magnitude of r.m.s. velocity fluctuations on the x - z plane (m/s)
R	Tube or artery radius (m or mm)
r	Radius (m or mm)
R_c	Characteristic resistance ($\text{kg}/(\text{m}^4\text{s})$)

R_p	Peripheral resistance (kg/(m ⁴ s))
Re	Reynolds number (dimensionless)
RS	turbulent (Reynolds) stress (kg/(ms ²) or Pa)
RS_{max}	Maximum turbulent stress (kg/(ms ²) or Pa)
T	Cycle period (s) or temperature (°C)
t	Time (s)
$t_{cut-off}$	Maximum number of standard deviations by which a value could differ from the mean without being considered an outlier (dimensionless)
U, V, W	Components of the velocity vector (m/s)
u, v, w	Velocity fluctuation components (m/s)
u', v', w'	Components of the r.m.s. velocity fluctuation (m/s)
V_{RC}	Velocity component in the axial direction of the right coronary artery (m/s)
W_0	Averaged axial velocity over a profile (m/s)
x, y, z	Components of the position vector in the test section (m or mm)
x^+, x^-, y^+, y^-	Positive and negative sides of the x and y axes, respectively (dimensionless)

Greek symbols

α	Womersley number (dimensionless)
α_M	Magnification factor (dimensionless)
α_η	Optical ratio (dimensionless)
α_A	Angle between artery axis and x -direction (radians or °)
α_N	Angle normal to artery axis (radians or °)
β	Lyne parameter (dimensionless)
$\Delta\tau$	Bin width of histogram (dimensionless)
Δt	Bin width for sorting single-point velocities (s)
δt	Time separation between PIV images (s or μ s)

ΔX	Particle displacement (pixels)
Δx_{lens}	Camera lens displacement (m)
Δx_{plane}	Displacement of the focus plane (m)
η	Refractive index (dimensionless)
λ_{ci}	Swirling strength (s^{-2})
μ	Dynamic viscosity of the fluid (Ns/m ² or cP)
ν	Kinematic viscosity of the fluid (m ² /s or cSt)
ρ	Density of the fluid (kg/m ³ or g/mL)
ρ_p	Density of particles (kg/m ³ or g/mL)
$\sigma_{max}, \sigma_{min}$	Maximum and minimum principal stresses (kg/(ms ²) or Pa)
τ	Viscous shear stress (kg/(ms ²) or Pa)
τ_{max}	Maximum viscous shear stress (kg/(ms ²) or Pa)
τ_P	Time constant of particle motion (s)
ω	Frequency of oscillation (rad/s)

Acronyms

2D	Two-dimensional
3D	Three-dimensional
BAF	Blood analog fluid
BDI	Blood damage index
BMHV	Bileaflet mechanical heart valve
DAQ	Data acquisition
DNS	Direct numerical simulation
LC	Left coronary sinus
LDV	Laser Doppler velocimetry
MCL	Mock circulation loop
MRI	Magnetic resonance imaging
NC	Non-coronary sinus

PDF	Probability density function
PIV	Particle image velocimetry
RC	Right coronary sinus
r.m.s.	Root mean square
SJM	Saint Jude Medical
STJ	Sino-tubular junction
TAVI	Transcatheter aortic valve implantation

Chapter 1

Introduction

1.1 Motivation

Heart disease is the leading cause of death globally, accounting for 30% of all deaths. One common form of heart disease is valvular disease. It was estimated that in 2009 more than 290,000 heart valve procedures were performed, and the demand for artificial heart valves was predicted to increase by 10-12% each subsequent year [28]. A diseased heart valve can be replaced with a bioprosthetic valve from animal tissue, commonly porcine or bovine, or with a mechanical valve made of man-made materials. Although bioprosthetic valves tend to have better hemodynamics (blood flow characteristics), their durability is inferior to that of mechanical heart valves. For this reason, mechanical heart valves are more commonly used in patients with long life expectancies and are implanted in more than half of all prosthetic valve recipients. The most commonly used mechanical heart valve is the bileaflet mechanical heart valve (BMHV), which houses two leaflets that pivot to create three flow channels, as shown in Figure 1.1. Despite the advances made to the design of mechanical valves over the past decades, there remain many problems associated with them: most notably, an increased risk of blood clot formation, known as thrombosis, and the subsequent dislodgement of the thrombus, referred to as an embolism. As a result, recipients are required to take life-long anticoagulant therapy, which causes, among other adverse side effects, an increased risk of hemorrhage.

In attempts to understand the mechanisms responsible for the observed problems and with the ultimate goal of improving valve designs, many studies, numerical and experimental, have examined the flow characteristics past mechanical heart valves. Research has identified non-physiological flow characteristics induced by BMHVs as



Figure 1.1: St. Jude Medical™ Regent™ bileaflet heart valve. (Reproduced from <http://professional.sjm.com/products/sh/mechanical-valves/aortic-mitral/regent#isw>)

one source of complications. High levels of viscous shear stress can (1) cause hemolysis, the destruction of red blood cells, which can result in an anemic condition or (2) cause platelets in the blood to become activated, which can initiate thrombosis [63, 61]. Regions of recirculation or slow flow in the wake of the valve can also increase the risk of thrombosis by causing prolonged contact between potentially activated platelets and proteins involved in coagulation, and by enabling thrombus attachment to the prosthesis or vessel wall.

The majority of *in vitro* and *in silico* studies of prosthetic valves have used very simplified geometries to model the heart and its vessels. The complex geometry of the aorta is commonly modelled as a straight tube, sometimes with an axisymmetric expansion to represent the Valsalva sinuses [85, 10, 14, 52, 36, 27, 11, 35, 84, 42, 88, 89], or having a simplified geometry of the sinuses without coronary arteries [6, 53, 38, 45, 4, 26, 74, 2, 41, 75]. In actuality, the aorta has a very complex geometry: it has three sinuses at its root from which two coronary arteries branch, and an arch that curves in three planes and branches to three additional large arteries (Figure 1.2).

A recent direct numerical simulation (DNS) study by Borazjani *et al.* [12] modelled the flow characteristics past a BMHV in an anatomically shaped aorta model. Despite the model being somewhat simplified, having had the branching arteries removed, it was found that the shape of geometry used at the exit of the valve had significant effects on the flow characteristics. The authors emphasized the need for a more anatomically accurate geometry, as well as experimental validation from which more

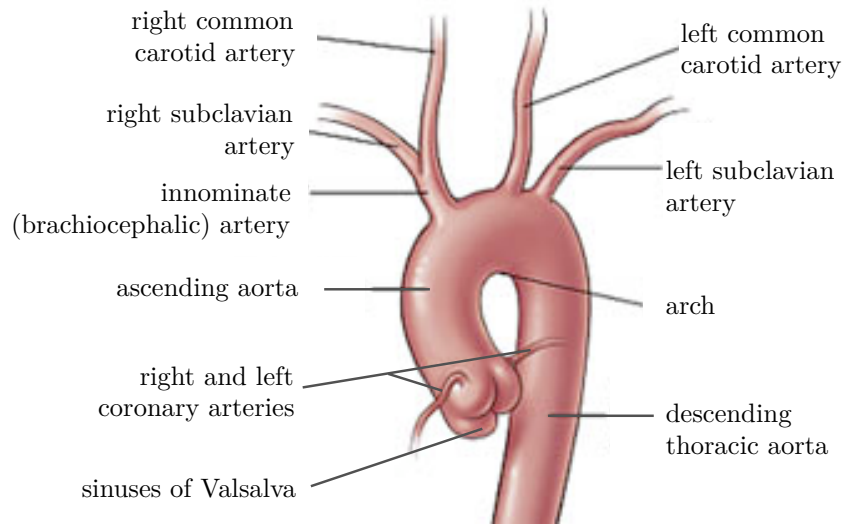


Figure 1.2: Diagram of the aorta with aortic root and sinuses, ascending aorta, descending aorta, and branching arteries. (Adapted from https://www.clevelandclinic.org/heartcenter/images/guide/heartworks/thoracoabdominal_aorta.jpg)

accurate upstream and downstream boundary conditions could be determined for use in future numerical studies.

Studies using a simplified sinus geometry without coronary arteries have reported recirculation zones within the sinus region. Although it is understood that these vortices assist native valve closure, their benefit towards BMHV kinematics is unclear. It is of great interest to understand the interaction of these vortical structures with the flow in the coronary arteries. Similarly, it has been inferred that flow through the arterial branches in the aortic arch may also affect BMHV flow characteristics. This configuration has not yet been studied experimentally or computationally. Of additional interest is how the azimuthal orientation of the BMHV within the sinuses affects the flow characteristics downstream of the valve. Understanding these effects could help to identify an optimal valve orientation for improved hemodynamics.

1.2 Objectives

The general objective of this research is to perform an *in vitro* study of the flow characteristics past a bileaflet mechanical heart valve (BMHV) under physiological flow conditions in (1) an axisymmetric channel and (2) an anatomical model of the aorta. The specific objectives of this research include the following.

1. To determine the velocity, viscous stresses, coherent structures and turbulence past the valve at various instances over the pulse cycle using laser Doppler velocimetry (LDV) and planar particle image velocimetry (PIV).
2. To compare the results from the two geometries in order to determine the effects of anatomical features including the sinuses, the coronary arteries and the aorta curvature.
3. To determine the effect of valve azimuthal orientation relative to the aorta from measurements made for three valve orientations within the anatomical model.
4. To investigate the effect of flow through the coronary arteries.

Viscous and turbulent stresses as well as regions of low flow and recirculation will be discussed regarding their implications toward thrombosis and other pathologies. It is expected that this study will provide a more accurate representation of the flow characteristics that occur in BMHVs *in vivo*. The results will contribute to a better understanding of the adverse hemodynamics caused by BMHVs and could be used towards a better valve design and an optimal implantation positioning that would improve the lives of its recipients.

1.3 Thesis organization

This thesis begins with a background and literature review in Chapter 2. The experimental apparatus and methods used are then described in Chapter 3, which also describes the post-processing procedures and the estimates of measurement uncertainties. The results are presented in Chapters 4, 5 and 6. First, Chapter 4 presents the measurements made in the axisymmetric channel: the first section describes the single-point measurements made with LDV, and the second presents the planar measurements made with PIV. The single-point and planar measurements are compared to each other and with previous measurements from literature. Chapter 5 presents the measurements made in the anatomical model of the aorta for one valve orientation and with no flow in the branching arteries. These results are compared with those from the axisymmetric channel, and differences are attributed to specific features of the aorta geometry. Chapter 6 presents the results for three different valve orientations in the anatomical model, first with no flow in the branching arteries, and then for flow through the right coronary artery. The Chapters 4, 5, and 6, are each concluded with an analysis and discussion of the chapter's results and their clinical implications. Finally, Chapter 7 summarizes the key findings and the contributions made, as well as the recommendations for future work.

Chapter 2

Analytical Background and Literature Review

2.1 Native heart valves: anatomy, kinematics and fluid dynamics

The human heart contains four valves that direct the flow of blood through the heart, allowing forward flow and preventing reverse flow. The pulmonic and tricuspid valves direct the blood on the right side of the heart that returns from the body and is pumped to the lungs to become oxygenated. The mitral and aortic valves direct the flow through the left side of the heart where the oxygenated blood from the lungs is pumped to the body. The pressure required to pump the blood to the entire body is clearly greater than that required to pump blood to the lungs, and as a result the valves on the left side of the heart are more prone to disease. The aortic valve, which faces the greatest pressures and velocities and is simpler than the mitral valve to model both experimentally and numerically, is the subject of many fluid dynamic studies and will be the focus of the following review.

The aortic valve is situated between the left ventricle and the aorta, the largest artery in the body. It is comprised of three semilunar cusps, or leaflets, that attach at the valve annulus and open from the valve centre, as shown in Figure 2.1. Immediately superior to the annulus is the aortic sinus, which comprises three bulges at the root of the aorta that accommodate the leaflets when they are open. These sinuses are sometimes referred to, collectively, as the sinus of Valsalva.

There are two phases of the cardiac cycle: systole occurs when the ventricle contracts and ejects blood into the aorta; diastole corresponds to the relaxation and filling of the ventricle. The aortic valve is closed during diastole, while the ventricle fills. When the ventricle begins to contract in systole, the ventricular pressure rises abruptly. When the ventricular pressure is greater than the pressure in the aorta, the aortic valve is forced to open, allowing blood flow into the aorta. A transvalvular

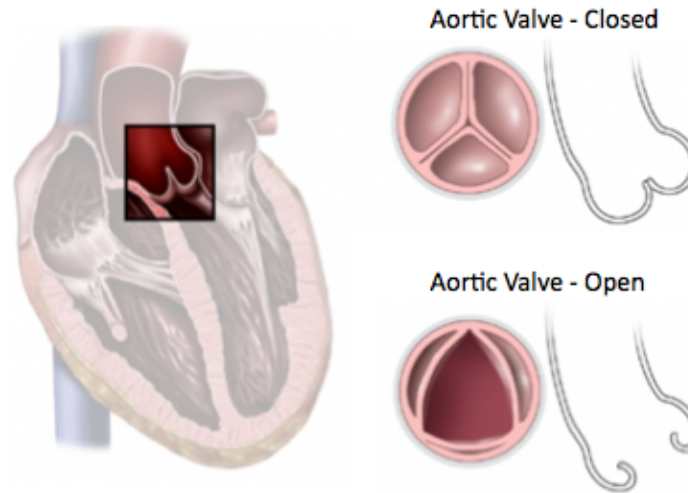


Figure 2.1: Diagram of aortic valve *in situ* (left) and cross-sections in closed (top right) and open (bottom right) positions. (Adapted from <http://heartsurgeryinfo.com/minimally-invasive-aortic-valve-surgery>)

pressure difference of only a few mmHg is required to open the valve [19]. The valve closes after the ventricular pressure decreases to a value below the aortic systemic pressure, and the flow decelerates to zero velocity. Typical systolic and diastolic aortic pressures are 120 and 80 mmHg, respectively. A diagram of the atrial, ventricular, and aortic pressures is shown in Figure 2.2. The aortic valve changes shape during the cardiac cycle with not only the leaflets opening and closing, but also with the valve annulus changing diameter and translating along the axis of the aorta.

The fluid dynamics through the valve are highly complex. Recirculation zones are formed in the sinuses behind the leaflets during systole, which are believed to help to close the leaflets at the end of systole to prevent back-flow into the ventricle. This observation was first made by Leonardo da Vinci in the early 1500's who, by dissection of cadavers, made detailed illustrations of the aortic valve and the sinus vortices as shown in Figure 2.3. It has been found from *in vitro* studies that the pressure gradient alone is sufficient to close the leaflets without existence of the sinuses, however, closure is not as quick as with vortices present [19]. The reverse flow volume in healthy people is found to be less than 5% of the forward flow volume. Other flow characteristics indicative of health are the velocity profile and peak velocities. The velocity profile

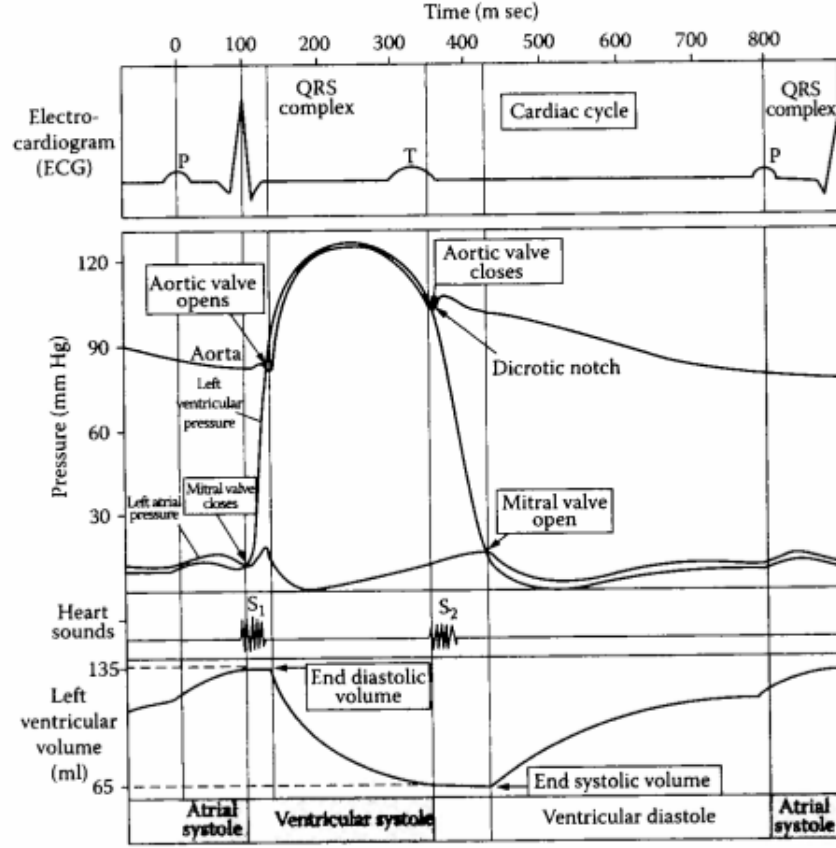


Figure 2.2: Wiggers diagram illustrating the aortic, ventricular, and atrial pressures, the left ventricular volume, and the electrocardiogram over the pulse cycle. (Reproduced from Chandran *et al.* [19])

at the annulus is normally quite flat, skewed slightly toward the septal wall (by less than 10% of centreline velocity). The velocity in healthy adults at peak systole can reach 1.35 ± 0.35 m/s, and can reach 1.5 ± 0.3 m/s in children [19]. In the presence of severe stenosis, velocities can reach values as high as 5 m/s and high levels of turbulent shear stress arise which may damage blood constituents [19].

2.2 Fluid dynamics in the aorta

One of the most significant geometrical features of the aorta is the fact that it has three planes of curvature. The primary plane of curvature enables the aorta to arch over the left pulmonary vessels and the left bronchus. The aorta curves again at the



Figure 2.3: Sketches by Leonardo da Vinci of the aortic valve (left) and the vortices in the sinuses of Valsalva (right). (Reproduced from <http://www.hmc.org.qa/hmc/heartviews/h-v-v3%20n2/8.htm>)

arch around the esophagus and trachea creating the second curve in the transverse plane. Additionally, the ascending aorta arches about the left ventricle in the third plane of curvature [21]. In order to understand the fluid dynamics within the aorta, it is critical to first have an understanding of the fluid dynamics in a simple curved tube.

Studies of flow in curved tubes were performed as early as 1911 when Eustice [33] found that streamlines moved towards the inner wall of a bend, and derived an equation for the resistance through a curved tube. Dean made further advancements with his theoretical analysis of fully developed steady flow in curved tubes. He discovered that a secondary flow develops at the tube cross-section due to an imbalance of centrifugal forces and the inward radial pressure gradient [21]. The fluid at the core moves toward the outer wall and then returns to the inner wall of the bend, creating two symmetric vortices, referred to as Dean vortices. The axial flow superimposed on these vortices results in a bihelical-structured flow with the maximum axial velocity shifted towards the outer wall of the bend. The magnitude of the secondary flow and

shift of maximum axial velocity is a function of the Dean number given by

$$De = Re\sqrt{\frac{R}{a}} \quad (2.1)$$

where Re is the Reynolds number, R is the tube radius, and a is the radius of curvature [21]. As a result of the secondary flow and the skewed axial velocity, a circumferential variation in the wall shear stress develops, with the maximum and minimum wall shear stresses occurring at the outer and inner walls respectively [21].

More relevant to the flow in the aorta is the study of pulsatile flow in curved tubes. The Womersley number α is a dimensionless parameter used to characterize periodic flows. It is defined as

$$\alpha = R\sqrt{\frac{\omega}{\nu}}, \quad (2.2)$$

where R is the pipe radius, ω is the frequency of oscillation, and ν is the kinematic viscosity of the fluid [19]. Lyne [55] studied the effect of a sinusoidally varying pressure gradient on flow through curved tubes. He identified an additional parameter

$$\beta = \sqrt{\frac{2\nu}{\omega R^2}} \quad (2.3)$$

which is the ratio of the Stokes layer thickness to the pipe radius R . Lyne found that for small values of β , the viscous effects are confined to the near wall region, whereas the inner core of the fluid may be modelled as inviscid. This results in an additional secondary flow structure, with a vortex pair near the wall rotating in opposite direction to the inner core vortex pair. The interior vortex pair rotates in the opposite direction than that predicted for steady flow in curved tubes. Chandran *et al.* [25] further showed in a numerical analysis that the vortices in the cross-section alternate over the pulse cycle from one pair to two pairs, as shown in Figure 2.4, in a complicated quadrihelical flow structure. Chandran & Yearwood [18] were the first to apply a physiological pulsatile flow to a curved tube as would occur in the natural aorta. This physiological pulse had a non-zero mean, superimposed with several harmonics of oscillatory flow. The study found that reverse flow was present along the inner wall of the curve during the diastolic phase of the cycle. In addition, they found that the maximum axial velocity was shifted towards the outer wall at the

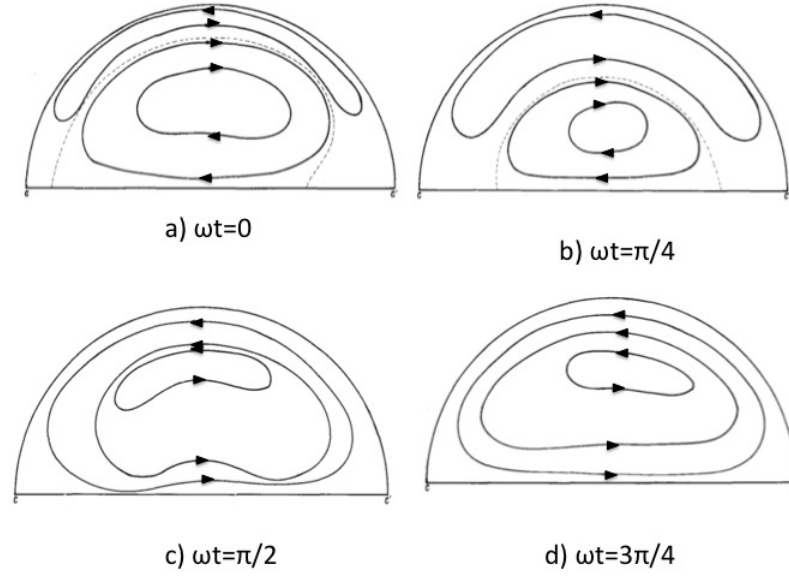


Figure 2.4: Secondary flow curves at the cross-section of a curved elastic tube at various instances over the cardiac cycle. At instances a) and b) there are two vortex pairs rotating in opposite directions - with axial velocity superimposed, the three-dimensional flow structure is quadrihelical. At instances c) and d) there is only one vortex pair - the flow structure is bi-helical. (Adapted from [25])

inlet, shifted towards the inner wall as the fluid progressed into the curve, and then back towards the outer wall downstream of the curve. They also observed trapped vortical motions at the inner wall in the downstream region.

Sudo *et al.* [68] studied experimentally and numerically the secondary flow structures in a curved tube under a laminar, oscillating flow (mean equal to zero) for varying Womersley and Dean numbers. They observed transitional flow morphologies and classified the secondary flow structures into five pattern types: (i) Dean circulation, (ii) deformed Dean circulation, (iii) intermediate circulation between Dean and Lyne, (iv) deformed Lyne circulation, and (v) Lyne circulation. Timité *et al.* [72] performed an experimental and numerical study of the secondary flow characteristics in a 90° bend, under pulsatile flow (mean not equal to zero) for varying Reynolds and Womersley numbers, and for two values of the ratio of maximum cycle velocity to mean cycle velocity. They found that, during flow acceleration, the intensity of secondary flow decreased with increasing Womersley number, but that during flow deceleration, the

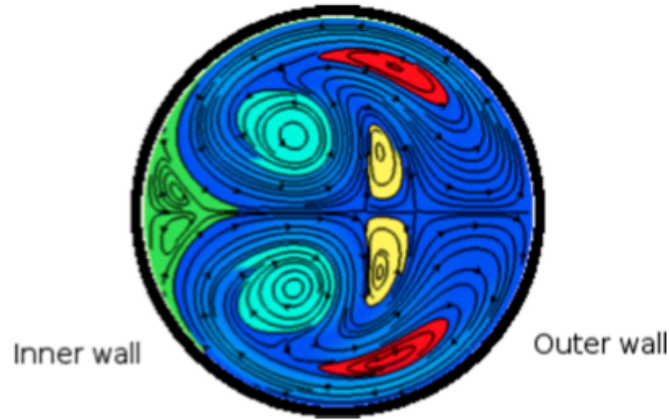


Figure 2.5: Four types of vortices identified by Bulusu & Plesniak [16] in the cross-section of a 180° bend, under a carotid, pulsatile flow rate: Dean-type (cyan); Lyne-type (yellow); wall-type (red), and innerwall-type (green). (Reproduced from [83])

intensity increased, along with the development of Lyne vortices and reverse flow at the inner wall. Bulusu & Plesniak [16] used PIV and continuous wavelet transforms to characterize the morphological changes of secondary flow structures in a 180° curved tube, subject to a carotid artery flow rate. They found that, during systolic deceleration, large-scale coherent secondary vortices broke down into multiple, disordered, low-circulation vortical structures, from which they elucidated new transitional morphologies. In addition to the previously documented Dean- and Lyne-type vortices, they observed two other types of structures named innerwall-type and wall-type vortices; the later type had low circulation and appeared intermittently throughout the measurements. The four observed vortex types are illustrated in Figure 2.5.

Various studies over the past three decades, both *in vivo* and *in vitro*, have examined the flow in a physiological geometry of the aorta. Endo *et al.* [32] performed a flow visualization study of the steady flow in a transparent model of a canine aortic arch. They observed a clockwise single helicoidal flow in the arch as well as the formation of vortices in the sinuses. Various *in vivo* experiments performed on dogs and horses yielded results consistent with the *in vitro* experiments previously mentioned. Using hot film anemometry and pulsed Doppler techniques, such studies found the velocity profile to be skewed towards the inner wall of curvature in the ascending

aorta, while being flat in the descending aorta during the systolic phase of the cardiac cycle. During diastole, however, flow reversal was found to develop at the inner wall in the ascending aorta, and propagated into the arch and the descending aorta. [21]

2.3 Optimization of hemodynamics

The main concerns regarding the performance of a mechanical heart valve are its hemodynamics, its durability, and the biological response to the valve. A designer's objectives in optimizing the hemodynamics include minimizing or eliminating the following:

- the pressure drop across the valve
- regurgitation (ideally eliminated)
- regions of stagnation and flow separation in the valve's vicinity
- the turbulence produced (ideally eliminated)
- regions of high local shear stress.

The pressure drop across the valve is a measure of the valve's efficiency. A large pressure drop results in a greater required ventricular pressure and therefore more work to be done by the ventricle. Regurgitation, the reverse flow through the valve, is another measure of the valve's efficiency. High levels of regurgitation reduce the cardiac output, requiring the heart to contract more to meet the body's demands.

Regions of slow flow, stagnation and flow separation should be avoided as they can facilitate thrombus formation, calcification, and tissue overgrowth by increasing particle residence times, resulting in prolonged contact between blood elements including potentially activated platelets and proteins involved in coagulation. Such regions also facilitate thrombus attachment due to the absence of flow disturbances.

High levels of turbulence should be avoided as they increase the mixing of blood elements and entrainment of adjacent flow, which can accelerate thrombotic reactions

[89]. It has been found that elevated levels of turbulent stress can damage platelets and red blood cells: platelets can be activated which can initiate thrombosis; and red blood cells can be destroyed, a process called hemolysis.

In addition to these general factors important to the design of prosthetic valves, two specific examples are discussed here:

- Bluestein *et al.* [10] were concerned about flow conditions that subjected cells first to high levels of viscous shear stress and then directed them to regions of low velocity where activated platelets could aggregate. Specifically, they identified the vortices shed from the leaflets as potential “flow-reactors”, in which platelets having just experienced the high shear levels surrounding the leaflets became trapped along-side other platelets and agonists. They found that any misalignment of the valve that resulted in a tilt angle to the flow direction enhanced this problem by causing stronger jets and larger wakes [9].
- The hinge mechanism of BMHVs has been heavily targeted as a feature that puts blood elements at risk. High velocity regurgitant jets squeezing through the small hinge orifice cause high levels of turbulent and viscous shear stresses, enhanced mixing, and areas of flow separation. A three-dimensional DNS by Simon *et al.* [64] characterized the flow around three hinge designs. They emphasized the importance of streamlined hinge geometries, avoiding sharp changes in geometry, to minimize flow separation, and maximize hinge washout.

2.4 Indicators of blood damage

As the most common problems associated with mechanical heart valves are hemolysis and thrombosis, it is important to understand the relationship between the mechanical forces acting on blood elements and the development of these problems. Under moderate levels of shear stress, red blood cells have been shown to change from the normal biconcave shape to an ellipsoidal shape. If the shear stress is sufficiently high, the cell will break open (lysis) and release its hemoglobin into the plasma. When this hemolysis is excessive, it results in an anemic-like condition. The damage to blood cells and platelets is a function of both magnitude of shear stress and the exposure

time to the stress. It has been found that there is a threshold stress level below which a cell will not be damaged. For red blood cells, this threshold has been found experimentally to be between 150 and 400 N/m^2 for long exposure times (on the order of minutes) [35]. The activation of platelets, which can initiate thrombosis, is more complicated than the lysis of red blood cells. Under normal circumstances, blood coagulates in response to an injured blood vessel to prevent hemorrhaging. When a platelet becomes activated (as a result of adhesion to exposed collagen, or its chemical or mechanical environment) it releases granules into its surroundings that contain potent agonist chemicals that amplify the activation and aggregation of other platelets and cells in a cascade reaction [51]. Platelets are much more sensitive to shear stress than red blood cells, and shear stress levels of 10 N/m^2 have been found to cause platelet activation [35].

Although much research has been made to determine the exact blood cell response to mechanical stress, there still remains much uncertainty and disagreement about this relationship. For many years, there was much ambiguity and confusion in reporting threshold levels of shear stress that damage red blood cells and activate platelets. Much of this confusion comes from the dual meaning of shear stress: in laminar flow only viscous shear stress exists, whereas in turbulent flow the term “shear stress” includes both the instantaneous viscous shear stress and components of the statistical stress-like quantity, the Reynolds turbulent stress. Ge *et al.* [35] argue that the turbulent stress is in fact a “pseudo stress” that is not representative of the true mechanical environment that affects blood elements, and that it is solely the viscous shear stress that causes damage to blood constituents. In contrast, it has also been argued that although turbulent stresses are a statistical quantity, they represent turbulent eddies which can release their energy onto blood cells, causing them damage or to be activated. Liu *et al.* [52] claim that if the smallest eddies (on the order of the Kolmogorov scale) are comparable to the size of the blood cells, they will interact directly with the cells and transfer their energy to them, which can damage the cell membranes. On the other hand, they affirm that if the smallest eddies are much larger than the diameter of the blood cell, they will completely surround the cells, rotating and translating with them without causing damage. Liu *et al.* [52]

studied the turbulence characteristics downstream of three BMHV designs within a pulse duplicator and measured the velocity fields in the sinus area using LDV. They found the lowest value in the smallest length scale to be $20\text{ }\mu\text{m}$, which is less than three times the $8\text{ }\mu\text{m}$ diameter of a red blood cell and five times the $4\text{ }\mu\text{m}$ diameter of platelets, which they believe is within the range of causing damage to cells.

In agreement with Ge *et al.*, numerous researchers have reported the viscous shear stress measured past a BMHV as the sole parameter relevant to blood damage [31, 12, 82, 43]. Other researchers have reported a combined stress, determined as the sum of the Reynolds shear stress and the viscous shear stress [9, 6, 84].

Although turbulent stresses are average properties and do not act directly on the blood elements, clinical and experimental evidence shows that turbulent stress level is better correlated with cell damage than viscous level, which is smaller by typically two or three orders of magnitude [44]. To explain this dichotomy, it was suggested by Antiga & Steinman [5] that although turbulent stresses do not physically act on cells, the associated turbulent fluctuations cause an increase in the local instantaneous viscous shear stress that they experience. They performed a simple order-of-magnitude calculation and estimated that the true viscous shear stress experienced by cells in a turbulent flow was intermediate between the laminar viscous shear stress and the turbulent stress.

Over the past three decades, there have been various blood damage models proposed to relate shear stress (viscous or turbulent) and exposure time to blood damage. To model hemolysis, Blackshear *et al.* [8] proposed a power-law model from the results of measuring the free hemoglobin released from blood exposed to constant shear in a Couette viscometer. It has since been acknowledged that this model does not account for time-dependant shear or the history of shear that accumulates over the life of a blood cell. Similarly, commonly used zonal approaches, which simply indicate regions of high turbulent or viscous shear stress, do not give any indication of exposure times or damage history of the blood cells [74].

De Tullio *et al.* [74] proposed using a Lagrangian approach for use with numerical simulations in which the cumulative damage is integrated over the path of a cell over the course of its lifetime. They calculated a maximum turbulent stress quantity

RS_{max} from the principal stresses of the Reynolds stress tensor to obtain a single value to represent the maximum turbulent stress. Similarly, a scalar equivalent of the viscous shear stress, τ_{max} , was calculated from the rate of strain tensor to represent the maximum viscous shear stress acting on blood cells. Passive particles were released into the fluid domain, and the instantaneous stresses that they encountered were integrated along their trajectory. De Tullio *et al.* [74] released and tracked 101 passive particles in their DNS of a BMHV in a physiologically shaped aortic root. Their results identified three very different trajectories typical of a cell: a cell could reside in regions of high-speed flows, away from any walls and pass through the valve very rapidly and experience negligible damage; while another could pass near the leaflet or housing wall and experience the corresponding shear stresses; while in a third trajectory, a cell could pass through very narrow gaps between the leaflets or the valve housing and experience large magnitudes of stress. It was reasoned that during the life of a red blood cell (half-life of 120 days) a cell would pass through the valve on average 1.73×10^5 times and that each cell would experience each trajectory type over its life. De Tullio *et al.* used the viscous shear stress to calculate a blood damage index (BDI), reasoning that Reynolds shear stresses cannot be accounted for on a particle trajectory of an unsteady flow because the turbulent stresses describe the average characteristics of the flow, while a particle trajectory is only affected by the local and instantaneous features of the field. This reasoning supports the theory of Ge *et al.* [35] that only viscous shear stress causes damage to blood cells. Following the method outlined by Grigioni *et al.* [38], they calculated the BDI by introducing the mechanical dose function, Ds , as:

$$Ds = t \cdot \tau^{\frac{c_2}{c_1}}, \quad (2.4)$$

where c_1 and c_2 are constants to be determined experimentally. A similar procedure was also used by Sirois & Sun [65], Shahriari *et al.* [62], and Forleo & Dasi [34].

Although such Lagrangian methods account for the time-history of the mechanically imposed viscous shear stress on blood, they are difficult to implement experimentally. Additionally, to be accurate, they rely on the precise calculation of the viscous shear stress field at a resolution that is comparable to the Kolmogorov scale;

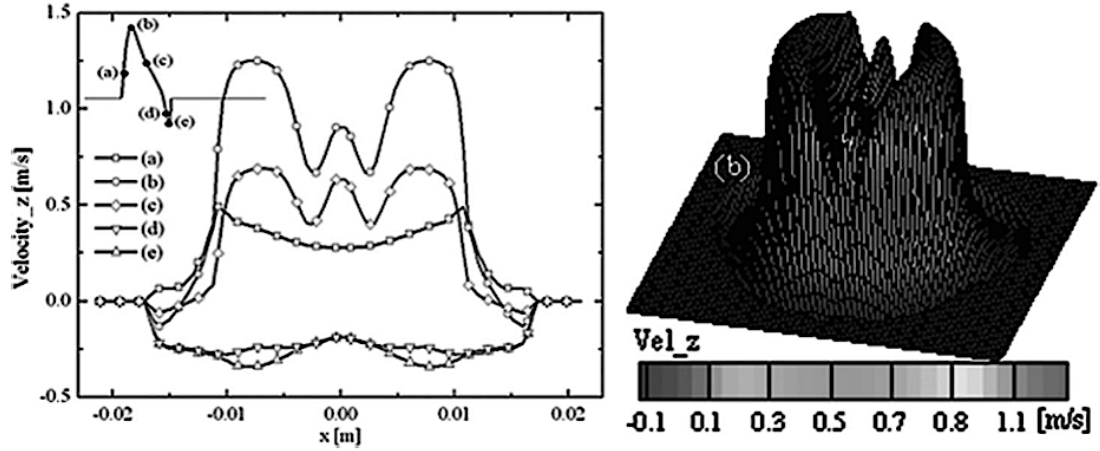


Figure 2.6: Velocity profiles and 3D flow reconstruction illustrating the triple jet of a BMHV. Reproduced from Choi & Kim [26].

such resolution is difficult to achieve in either experimental or numerical studies, because it is also necessary to resolve large-scale motions [89]. In conclusion, it seems appropriate in this experimental study to examine the distributions of both viscous shear stresses and turbulent stresses as indicators of relative risk to blood integrity. Additionally, regions of pathological stagnation, or persisting low flow should be identified as regions that could promote thrombus growth or attachment.

2.5 General fluid dynamics of BMHVs

The flow past a BMHV is characterized by a triple jet structure corresponding to the three orifices of the valve (Figure 2.6). The bulk of the flow is through the two lateral channels that have peak jet velocities of 2.2 m/s, while the central jet reaches velocities of 2 m/s at peak systole. The peak velocity is reached at 8 mm downstream of the valve, and is maintained to a distance of 13 mm. At a distance of 20 or 30 mm downstream of the valve, the three flow jets merge into a central flow region. Regions of flow separation exist at the boundaries of the jets, adjacent to the wall of the flow channel. Regions of reverse flow exist adjacent to the valve hinge. High levels of turbulent shear stress exist at regions of high velocity gradient, notably immediately distal to the leaflets [86].

A numerical study by Choi & Kim [26] characterized the leaflet dynamics of a BMHV in a triple sinus aortic root under physiological flow conditions. They found that at the beginning of systole, the opening velocity of the leaflets quickly increased to 47.4 rad/s and then diminished until the valve is fully open. During the closing phase, however, the leaflet closing velocity exponentially increased in the opposite direction until reaching 148.9 rad/s at impact, followed by the rebound of the leaflet due to its inertia moment and the water hammer effect. They observed that this elevated closing velocity resulted in squeeze flows between the leaflets and the valve housing, which were associated with very high shear stresses. They also noted that the reverse flow during closing caused a low pressure to develop on the ventricular side of the leaflets which could result in cavitation.

Dasi *et al.* [27] performed an experimental and numerical study of the vortex dynamics of a BMHV in an axisymmetric aortic root, elucidating the complex three-dimensional vortex structures over the pulse cycle. PIV was used to measure an ensemble of instantaneous two-dimensional velocity fields in the vicinity of a St. Jude Medical 23 mm Regent BMHV. A direct numerical simulation (DNS) was performed in two dimensions to compare against the 2-D experimental results, and then in three dimensions to capture the 3-D evolution of the BMHV flow structures. The DNS accounted for the motion of the leaflets, and their kinematics were prescribed from the experimental measurements. The DNS did not, however, account for the non-linear fluid structure interaction of the aorta, and modelled it as a rigid structure. The study's description of the flow characteristics, beginning with the findings of the 2-D results, is summarized in the following.

The flow field about the BMHV at the beginning of systole was characterized by various randomly spaced, weak vortical structures. These were the decaying remnants of the previous cycle, and the ensemble-averaged field was void of vortical structures. At the onset of flow acceleration, these remnant structures were rapidly washed away. The two leaflets were pushed open, creating the triple-orifice flow previously documented [86]. The two lateral jets were bounded by two shear layers: one originating from the separation of the valve housing, and another from the separation at the leaflet tip. [27]

The shear layer from the housing separation rolled up as a ring-like structure into the sinus region, referred to subsequently as the “sinus vortex”. It was observed on closer examination that this ring extracted a secondary layer of vorticity at the sinus wall, of opposite direction to the first. As acceleration progressed, the sinus vortex travelled downstream into the end of the sinus region and grew in strength and complexity. A third vortex layer at the sinus wall was formed with the vorticity of the primary, secondary and tertiary layers being negative, positive, and negative, respectively. These observations corresponded with the findings of other studies of the flow of a vortex ring impinging on a flat wall. This suggested that the observed vorticity-wall interactions were inherent characteristics of a pulsed vorticity ring advancing into a sinus-like contraction. [27]

In the same manner as for the housing separation, shear layers were generated on either side of each leaflet, the outer of which rolled up into a ring-like structure, subsequently referred to as the “start-up vortex”. This ring travelled downstream much faster than the sinus vortex. As acceleration progressed, von Kármán vortex structures developed in the region between the start-up vortex and the leaflets. These structures were advected downstream in the wake of the leaflets, and formed two complex vortex layers that interacted with the shear layer in the sinuses. [27]

At the early stages of systole, the flow characteristics were highly organized and repeatable from cycle to cycle. The characteristics became non-repeatable when the shear layers of the valve leaflets began to shed von Kármán vortices. Just past the peak of systole, and at the onset of deceleration, the organized von Kármán vortices were broken up into chaotic, fine-structured turbulence. Despite the viscous dissipation of vortical strength, this chaotic turbulence continued to dominate the flow even after the closure of the valve, and remained until washed out by the incoming flow of the subsequent cycle. Significant cycle-to-cycle variations of the leaflet kinematics in the deceleration phase were attributed to the randomness of turbulence. [27]

The 2-D results from the DNS were in good agreement with the experimental results, indicating that the general flow structure, characteristic of the BMHV, is only weakly dependant on the exact hinge geometry, inlet flow conditions, and the cycle-to-cycle variations in kinematics. It was concluded that the essential flow physics

were governed by the large-scale geometry of the valve and by the pulsatile nature of the flow. [27]

In the same study by Dasi *et al.* [27], three-dimensional direct numerical simulation (DNS) uncovered the evolution of the 3-D flow structures downstream of the valve. These simulations found the sinus vortex to be a dominant feature in the flow, which is shown as *VR2* in Figure 2.7. Due to the asymmetry of the valve and consequential higher velocity in the lateral orifices, the vortex ring was advected downstream non-uniformly, and bent into a V-shape as illustrated in Figure 2.7(c). Insight to the evolution of the leaf-tip “start-up” vortices was also achieved. They appeared at the start of systole as tube-like structures that took the shape of the leaflet edge to which they were attached, shown in Figure 2.7(a) as *VT1* and *VT2*. As acceleration progressed, they lifted off the leaflets and were advected up into the sinuses as shown in Figure 2.7(b). The leaflet start-up vortices attached to the sinus vortex ring, and continued to stretch downstream, developing into hairpin-like vortices that led the entire complex vortex structure, which can be seen in Figure 2.7(e). A rectangular vortex ring was found to originate from the central orifice jet, shown as *CJ* in Figures 2.7(c) and 2.7(d). The two short sides of the ring travelled faster than the long sides, resulting in the bending of the ring. The rectangular ring was connected to the large sinus ring in the sinuses by two pairs of streamwise vortices. At the onset of deceleration, the complex vortex structure disintegrated in an “explosive manner” into fine, chaotic turbulence, as illustrated in Figures 2.7(f) and 2.7(g) [27].

2.6 The effects of physiological geometry

The work by Dasi *et al.* [27] made significant advances in our understanding of the flow field induced by BMHVs. Their findings, however, have limited relevance to physiological conditions, as the simulations were performed in a straight, axisymmetric aorta model. Chandran *et al.* used flow visualization and laser Doppler velocimetry to study the pulsatile flow through a bioprosthetic valve [22], a cage and ball valve [23], and a tilting disk valve [24] at the entrance of a model aorta. Using LDV, they obtained the one-dimensional velocity profiles at four cross-sections of the aorta and

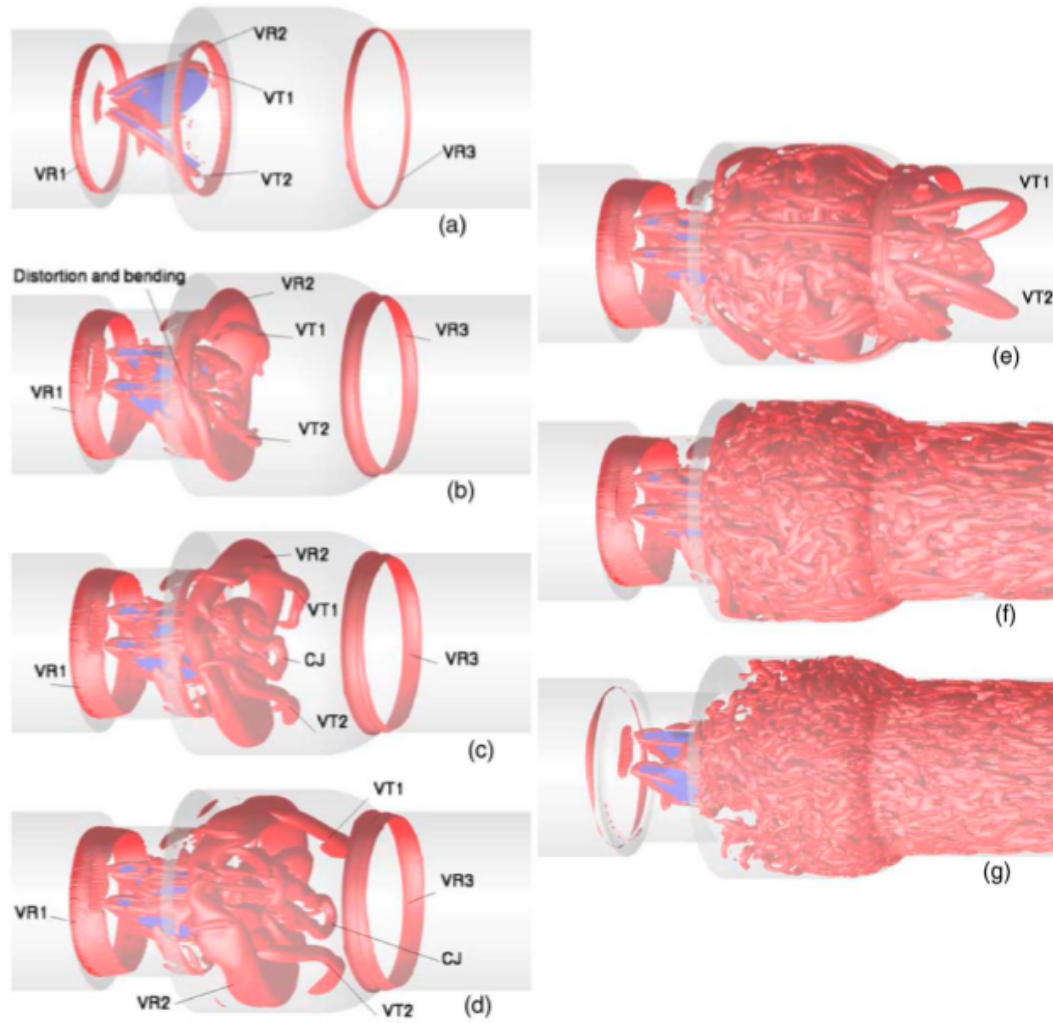


Figure 2.7: Three-dimensional coherent structure evolution defined by Q-criterion over the pulse cycle. (a) opening phase, (b)-(d) progressive acceleration, (e) peak systole (f) peak systole (g) deceleration. (Reproduced from [27])

calculated the turbulent stresses downstream of the valves, and the pressure drop across the valves. The results indicated that the flow characteristics downstream of the valve depended on the valve type as far as the middle of the aortic arch, downstream of which, into the descending aorta, the valve type had no effect. The study found that the bioprosthetic valve had superior hemodynamics over the cage and ball and tilting disc designs.

Borazjani *et al.* [12] performed the first DNS study of the flow characteristics induced by a BMHV in pulsatile flow in an anatomically realistic aorta. Their model was adapted from patient specific magnetic resonance imaging (MRI) scans; the branching arteries including the coronary arteries were removed, and a BMHV was placed at the inlet of the sinuses. The model geometry, shown in Figure 2.8, was adapted to have straight sections upstream of the valve, as well as at the exit of the descending aorta.

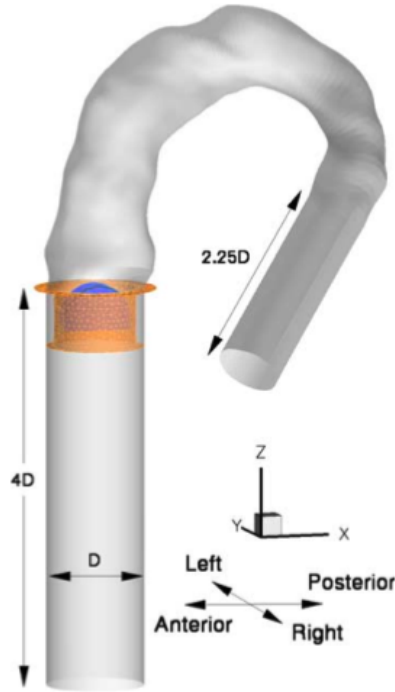


Figure 2.8: Anatomically realistic model used by Borazjani *et al.* [12] with branching arteries removed, and adapted straight sections at inlet and outlet, mounted with a BMHV.

The 3-D anatomical results were compared with those obtained by the study

of Dasi *et al.* [27], which used the straight aortic geometry. A comparison of the leaflet kinematics demonstrated that the leaflets in the anatomic aorta opened faster and experienced a greater rebound in closing than in the straight aorta. The flow in the anatomic model was characterized by the large scale features present in the axisymmetric model including the triple-jet structure, the sinus vortex in the sinuses, and the existence of shear layers on either side of the valve leaflets. The sinus vortex was also found to extract a secondary vortex layer of opposite sign at the sinus wall, the change in vorticity direction indicating the point of reattachment of the lateral jet. [12]

It was found that, due to the smaller expansion geometry of the anatomic aorta, the lateral jets reattached further upstream than in the case of the straight aorta. It was also found that the curvature of the anatomic aorta resulted in secondary motions (as described in Section 2.2), which prevented the jets from penetrating downstream. These two features resulted in greater lateral diffusion of the jets and lower jet velocities. As a result of this lateral diffusion of the leaflet shear layer, it did not roll up into the start-up vortex as it did in the straight aorta. As a result of the lower jet velocities, the leaflet shear layers remained stable longer in the anatomic model prior to breaking down into von Kármán vortex structures. Just prior to peak systole, the laminar, organized vortex structures broke down into small chaotic turbulent structures in the same manner as in the straight model. [12]

The 3-D results captured additional discrepancies between the anatomic and straight models. Vortex structures were found to form in the aortic arch of the anatomic model at early systole. They were resultants of secondary flow currents due to the curvature of the arch and were non-existent in the straight model. Additionally, the non-axisymmetry of the sinuses affected the shape evolution of the sinus vortex ring. Whereas in the straight section it was found to bend in a V-shape, in the anatomic model it deformed into a trimodal geometry due to the triple sinus structure. Borazjani *et al.* [12] made an assessment of the hemodynamic forces applied to blood elements using the metric of viscous forces as recommended by Ge *et al.* [35]. They found that regions of maximum local shear stress were more widespread in the anatomic aorta due its curvature, and were most significantly located near the leaflets

and in the aortic arch.

The study of Borazjani *et al.* [12] demonstrated the critical effect of aortic geometry on the flow characteristics in BMHVs and emphasized the need for future work considering compliant vascular walls, ventricular geometry, physiological upstream and downstream conditions, blood flow distribution in aortic branching arteries, and the effects of non-Newtonian fluids.

Other studies that highlight the effects of physiological geometry on BMHV flows include those of de Tullio *et al.* [74, 73, 75], and Grigioni *et al.* [37]. In their 2009 study, in which they performed a 3D DNS of a BMHV in a physiological shaped sinus, de Tullio *et al.* [74] made observations on how the triple sinus structure of the physiological aortic root affects the leaflet dynamics, primarily as a result of its non-axisymmetry. They observed that the shear layers that were released off the leaflet tips had a strong flapping motion, which contributed to the unsteadiness of the flow, and that this effect was the greatest for the leaflet that opened most directly into a sinus, as a result of the interaction of its wake with the sinus recirculation. They also observed that the non-axisymmetry of the sinuses resulted in an asymmetrical pressure field which caused an asymmetry in leaflet closure. This was a direct result of the recirculation in the sinus during diastole that produced a low pressure distribution on the aortic side of the leaflet causing the delayed closure of the leaflet that it was aligned with. In their three-dimensional FSI study in which they compared three aortic root geometries (straight tube, circular sinus, and triple sinus), de Tullio *et al.* [75] found that the angular velocities of the leaflets immediately prior to closure impact were significantly reduced in the physiologic-like triple sinus, compared with the circular or straight tube geometry. This has important implications regarding valve performance, considering that high closing velocities are related to hemolysis, in which red blood cells are squashed.

As previously mentioned, an important feature of the sinus of Valsalva are the vortices that form within them. The native sinus presents a clear evolutionary advantage in conjunction with the native aortic valve, as it provides three recesses for the leaflets to open into during systole, with three corresponding vortices that form during diastole that help to push the leaflets closed. Although it is still uncertain

whether the triple sinus geometry is advantageous when the native valve is replaced with a bileaflet valve, it is certain that the flow characteristics past the valve are influenced by this root geometry.

In their three-dimensional numerical simulation, Grigioni *et al.* [37] compared the flow characteristics of a BMHV in a physiologically shaped sinus (modelled by epitrochoid functions) to a straight tube geometry. Not surprisingly, they found that the physiological root developed larger vortices in its sinus compared with the corresponding regions in the straight tube. They projected that this was a favourable condition for coronary flow. They commented that the spatial progression of these vortices was highly dependant on the sinus geometry and that using a physiologically shaped sinus provided a better flow representation for both numerical and experimental simulations. Their results also indicated higher axial velocities of the triple jets and greater forward flow for the physiological geometry compared with the straight tube. They also noted that the “trailing vortices off the leaflets” (i.e. the start-up vortices) appeared to have specular vortices that resided in the sinuses (presumably the “sinus vortex”) which were not present in the straight tube model; however, they made no speculation on the implications of this observation. In agreement with the results of de Tullio *et al.* [74], Grigioni *et al.* reported that the triple sinus geometry resulted in a non-uniform pressure field, and that the pressure fell immediately downstream of the leaflets and partially recovered toward the end of the aortic root. As a result of these effects, Grigioni *et al.* reported that the transvalvular pressure drop at peak systole was underestimated by 14% when using a straight duct model as compared with the physiological root. Despite these differences, Grigioni *et al.* found little variation in the shear rate distribution between geometries, with the exception for a region of low shear rate within the sinuses (seen in the transversal plane) that could imply a lower stress acting on blood elements.

Although de Tullio *et al.* [75] reported that the geometry of the sinuses had minimal effect on the gross flow kinematics of the valve, they emphasized that the geometry had a significant effect on the pressure field. Additionally, de Tullio *et al.* [74] observed strong retrograde jets within the sinuses during diastole as a result of the reverse current along the wall, which they speculated would have an important

effect on coronary flows. Despite this hypothesis, in their 2011 study, in which they compared the effect of root geometry on coronary flows [75], de Tullio *et al.* reported that the Valsalva sinus had minimal effect on coronary entry flow.

2.7 The effects of valve orientation

Borazjani & Sotiropoulos [13] extended their numerical simulation of a BMHV in an anatomical aorta to investigate the effect of valve orientation. They assessed the leaflet kinematics and the fields of vorticity, pressure, and viscous shear stress for three different valve orientations: the “0 degree” orientation, in which the valve’s plane of symmetry through the central orifice¹ is parallel to the plane of curvature of the ascending aorta; the “90 degree” orientation which is perpendicular to the “0 degree” orientation; and the “45 degree” orientation, which is half-way between the other two. They found that valve orientation significantly affected the symmetry of leaflet closure, with leaflets closing nearly simultaneously in the “0 degree” orientation (1 ms time difference between closure of the two leaflets), but with increasing asymmetry for increasing orientation angle (41, and 76 ms for the “45 degree” and “90 degree” orientations, respectively). They explained this response as the result of imbalances in the pressure field caused by the aorta curvature. For the “0 degree” orientation, in which the valve symmetry aligned with the ascending aorta plane of curvature, the pressure field was symmetric with respect to the leaflets and resulted in a symmetric closure. For the other orientations, however, the pressure field was not symmetric with respect to the leaflets, and was greatest towards the outside of the bend in the aorta (a characteristic of all pipe bend flows). As a result, the leaflet positioned nearest the outer wall closed before the other.

The results of the study also indicated that leaflet rebound was the least for the “0 degree” orientation, and increased with increasing orientation angle. Observing that asymmetric valve closure corresponded to increased valve regurgitation, and large leaflet rebounds can result in cavitation induced wear and hemolysis, Borazjani & Sotiropoulos [13] suggested that the ideal orientation of a BMHV is such that its

¹Referred to subsequently as simply “the valve’s plane of symmetry”.

plane of symmetry corresponds with the plane of curvature of the ascending aorta. Additionally, they found that orientation had negligible effect on the magnitude or distribution of viscous shear stresses, however, they did find that orientation affected the evolution of vorticity within the aorta. In particular, they observed that the vortex ring, which evolved into a tri-modal shape for the 0 and 90 degree orientations (corresponding with the findings of their previous study), developed into a V-shape for the 45 degree orientation. [13]

In their 2008 and 2010 studies, Akutsu *et al.* [4, 2], experimentally investigated the effect of BMHV orientation within an anatomic-like sinus in a pulse duplicator. The sinus consisted of three cusps, without coronary ostia, and returned to a straight tube at the end of the cusps. Using dynamic particle image velocimetry (DPIV), they examined two orientations each for three BMHV designs: the Saint Jude Medical (SJM), the On-X, and the Edwards MIRA valve. The orientations examined were the “180°” orientation, in which the valve’s plane of symmetry bisected the non-coronary cusp, and the “90°” orientation, which was perpendicular to the prior. They reported that, for all three valves, the flow was stronger and less diffuse when the valve was in the “180°” orientation. Also in this orientation, the forward flow was more simple and central, and the washout in the sinus was stronger compared to the other orientation.

In a subsequent study, Akutsu *et al.* [3] measured the velocity past the same three BMHV models in a channel with a triple sinus and with outlets representing the coronary arteries. They measured the velocity in the coronary arteries with an ultrasonic flow meter and found that, for all valves, the coronary flow rate correlated with the strength of recirculating flow in the sinuses; for two of the three models (including the SJM) the coronary flow was higher when the valve’s plane of symmetry was aligned with one of the three sinuses compared to when it was rotated 90° from that.

In vivo animal studies have also been used to study the effects of bileaflet valve orientation. In their 1998 study, Kleine *et al.* [47] implanted SJM bileaflet and Medtronic Hall tilting disc valves into four healthy pigs (each valve model in two pigs), using a device that allowed rotation of the valve without reopening the aorta. They assessed the effect of valve orientation on turbulence downstream of the valve by measuring the

Reynolds normal stress, using pulsed Doppler ultrasonography. They found that the tilting disc valve showed significantly lower levels of turbulence when the major orifice was oriented towards the right-posterior aortic wall, which experienced the highest velocities during ejection. The SJM bileaflet valve also had lower turbulence when one orifice was oriented towards this region, however the reduction in turbulence was not as great.

In their 2002 study [48], they repeated the experiment, but investigated the effect of orientation on left coronary flow and, in addition to the Medtronic Hall and SJM valves, also tested the Medtronic Advantage bileaflet valve. Flow in the left anterior descending coronary artery was measured for each valve for two orientations, defined in a previous study as the best and worst hemodynamic positions [46]. It was found that, for the bileaflet valves, the left coronary flow was higher when one of the lateral orifices was aligned with the right coronary cusp, compared to the orientation perpendicular to it. This finding indicated that, because the majority of coronary flow occurs during diastole, the valve orientation affected the downstream flow field even during diastole when the valve was closed. The authors reported that the coronary flow rate was less sensitive to valve orientation for the SJM valve model, compared to the other two models.

Figure 2.9 summarizes the BMHV orientations recommended from these studies (the coloured lines indicate the lines of valve symmetry). This illustrates that, while each study indicated that BMHV orientation does indeed affect certain aspects of the flow past it, the orientations recommended for optimizing various parameters varied widely. The best orientation for coronary flow found by Akutsu *et al.* [3] was close to the one recommended by Kleine *et al.* [48]; i.e. with one lateral orifice facing the right coronary cusp, or the line of symmetry intersecting the non-coronary cusp; this was very different from the orientation found by Kleine *et al.* [47] for reducing turbulence, in which one of the valve’s lateral orifices faced the right-posterior wall. While Borazjani & Sotiropoulos [13] found that valve orientation had negligible effect on viscous stresses past the valve, they recommended that the valve be oriented in-line with the direction of aorta curvature for more synchronous valve kinematics and potentially less regurgitant flow, adding another “best” orientation to our list.

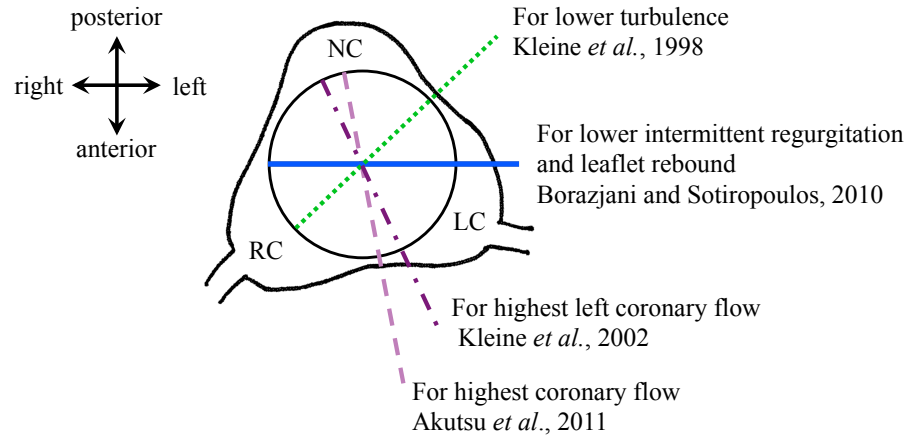


Figure 2.9: BMHV orientations recommended by previous studies, relative to the aortic root geometry (indicated by lines of valve symmetry through its central orifice).

As these results suggest, it may be that different valve orientations are required for optimizing different parameters; e.g., one orientation may be best for reducing fluid stresses, while another may be best for increasing coronary flow. In this case, a surgeon would be required to choose the best orientation for each patient's unique condition.

Each of the studies discussed here had inherent limitations. The model of Akutsu *et al.* [3] did not include the aorta's curvature; to my knowledge, there has not yet been an experimental study of a BMHV within a fully anatomical model of the aorta. Measurements of turbulent stresses made by Kleine *et al.* [47] were limited to several discrete points. That study, as well as their one of coronary flow in 2002, used only two animals for each valve and had no measure of the flow patterns inside the aorta to explain the results. The model of Borazjani & Sotiropoulos [13] did not include the coronary arteries and coronary flow.

This thesis research addresses many of these limitations: the effect of BMHV orientation was studied experimentally within a fully anatomical model of the aorta within controlled and repeatable physiological flow conditions; the effects of valve orientation on Reynolds stress, viscous stress and coronary flow were determined from 2D PIV measurements made on various planes in two orientations; in Chapter 6 the flow patterns past the valve are examined and used to give a physical explanation of the results.

Chapter 3

Experimental Apparatus, Instrumentation and Procedures

3.1 Apparatus

3.1.1 Mock circulation loop

A mock circulation loop (MCL) was designed, built, and tested so that it closely reproduced the physiological flow environment of the human circulation system (Figure 3.1). A bileaflet mechanical heart valve (BMHV) was mounted within this apparatus, and velocity measurements were made in its vicinity using LDV and PIV.

The pulsatile flow of the MCL was driven by the compression of two flexible bulbs with compressed air. LabVIEW was used to control the pulse frequency and the duration of systole and diastole by operating a three-way solenoid valve that directed the compressed air to the flexible bulbs; signals between the computer and the solenoid valve were communicated via a data acquisition device (DAQ) (USB-6009; National Instruments, Austin, TX, USA). The fluid was supplied to the bulb from a fluid reservoir. Check valves located upstream of the bulbs prevented reverse flow and, consequently, the flow was forced forward when the bulb was compressed.

The fluid was pulsed from the ventricle bulb into the upstream compliance chamber C_1 which simulated the compliance of the aorta. The chamber was partially filled with the working fluid and contained a volume of air that was adjusted through a valved vent in the top of the chamber. The bottom portion of the chamber was filled with a filter material to eliminate jet and vortex structures produced by the bulbs. Next, the fluid entered the plenum which was internally shaped as an axisymmetric contraction, and accelerated the fluid into the test section. Pressure transducers (UNIK 5000; GE, Fairfield, CT, USA) were located both upstream and downstream of the BMHV; the pressures at these locations were denoted as P_v and P_a , respectively.

A second compliance chamber C_2 was located immediately after the test section

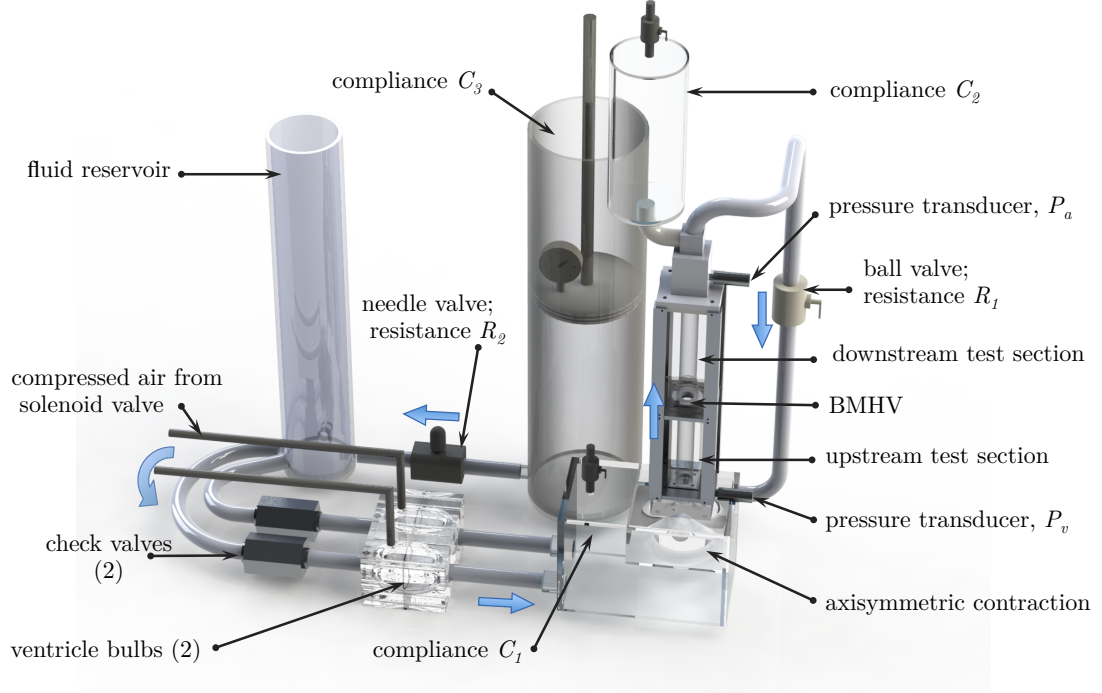


Figure 3.1: Computer model of the mock circulation loop.

to provide additional control of the pressure waveform. A ball valve was located downstream of this chamber to simulate resistance R_1 , which was analogous to the “characteristic resistance” R_c of the three-element Windkessel model of the circulatory system [79, 80] (Figure 3.2). Next, a third compliance chamber C_3 allowed the pressurization of the system. This chamber was originally intended to be the sole compliance chamber of the loop, but was located too far from the test section to have the desired effect. It was comprised of a large sealed cylinder filled partially with the working fluid and with an adjustable volume of air that was pressurized. Following this chamber, a needle valve simulated the resistance R_2 (analogous to the peripheral resistance R_p of the three-element Windkessel model) and regulated the return flow to the fluid reservoir.

3.1.2 Test sections

For initial testing of the MCL, a preliminary test section consisting of a cylindrical tube machined into an acrylic box with flat sides was used.

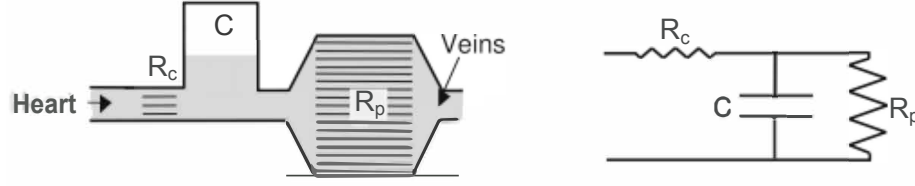


Figure 3.2: Hydraulic (left) and electrical (right) representations of the three-element Windkessel model of the circulatory system. (Adapted from Westerhof *et al.* [80].)

Following the testing and finalization of the loop, this section was replaced with an axisymmetric model that was cast out of rigid, clear silicone (Sylgard 184; Dow Corning, Midland, MI, USA) (Figure 3.3). This model consisted of an upstream section and a downstream section; the BMHV was mounted between them within a valve holder machined out of stainless steel. The upstream test section consisted of a cylindrical channel, with diameter $d_i = 25.4$ mm and 140 mm long. Including flanges and other connections, the straight channel upstream of the valve was 180 mm long. This was long enough so that, for stroke volumes of up to 90 mL, all of the fluid that passed through the BMHV during each cycle was inside of the straight channel at the start of the cycle, and effects of other upstream geometries would therefore have been negligible. The inlet of the downstream section was 31.8 mm in diameter and reduced to 25.4 mm after an axial distance of 20.3 mm. This expanded region was the axisymmetric representation of the three-dimensional aortic sinus, and will be referred to as the “pseudo-sinus”. This geometry, shown in Figure 3.3, was chosen to match that used in the experiments of Dasi *et al.* [27], in which the vortex dynamics of BMHVs were investigated experimentally and numerically.

Following completion of the measurements in the axisymmetric geometry, the downstream section was replaced with the anatomical model of the aorta. The aorta model was obtained from Elastrat (Geneva, Switzerland, upper half of model TRN-020), a company that specializes in fabricating vascular models for medical training. The model, shown in Figure 3.4, was cast out of the same silicone as was the axisymmetric model and was completely transparent, with no air bubbles or defects. It comprised the geometry of the triple sinus at the aortic root, the coronary arteries, the aortic arch with the branches of the brachiocephalic, left carotid, and left

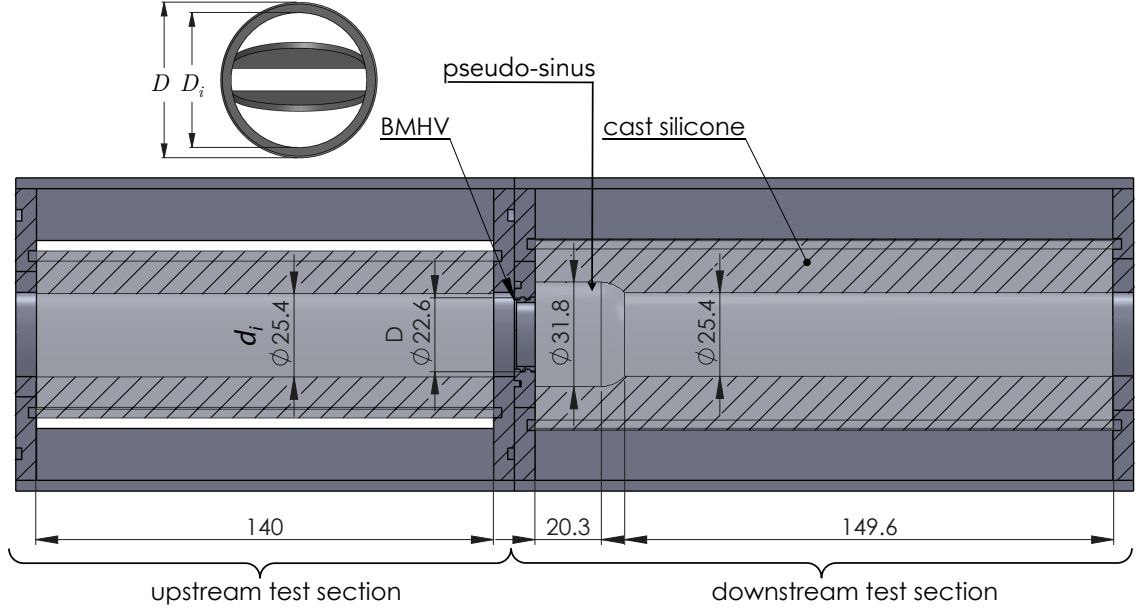


Figure 3.3: Axisymmetric, straight test section with sections upstream and downstream of valve. Geometry similar to that used by Dasi *et al.* [27]

subclavian arteries, and the upper part of the descending thoracic aorta. It had the “classical” or “normal” branching pattern of the aortic arch which occurs in between 65 and 94% of people [50]. Information about the individual from which the geometry was obtained is not known. The silicone model was mounted on a custom base-plate machined out of stainless steel, and encased in glass plates on the top and side walls. The outlets from the arteries branching off of the aorta were fitted with adjustable clamps in order to control the flow through them. The arterial outlets reconnected to the main flow downstream of the model via a single return hose.

3.1.3 Working fluid

The working fluid of the MCL was a mixture of water (47.38% by weight), glycerin (36.94%), and sodium iodide (15.68%). This mixture was proposed by Yousif *et al.* [87] to approximately match the viscosity of blood at physiological shear rates and to match the refractive index of clear silicone. By matching the fluid’s refractive index to that of the test section walls (which are made of clear silicone), undistorted optical access for PIV and LDV measurements was effectuated.

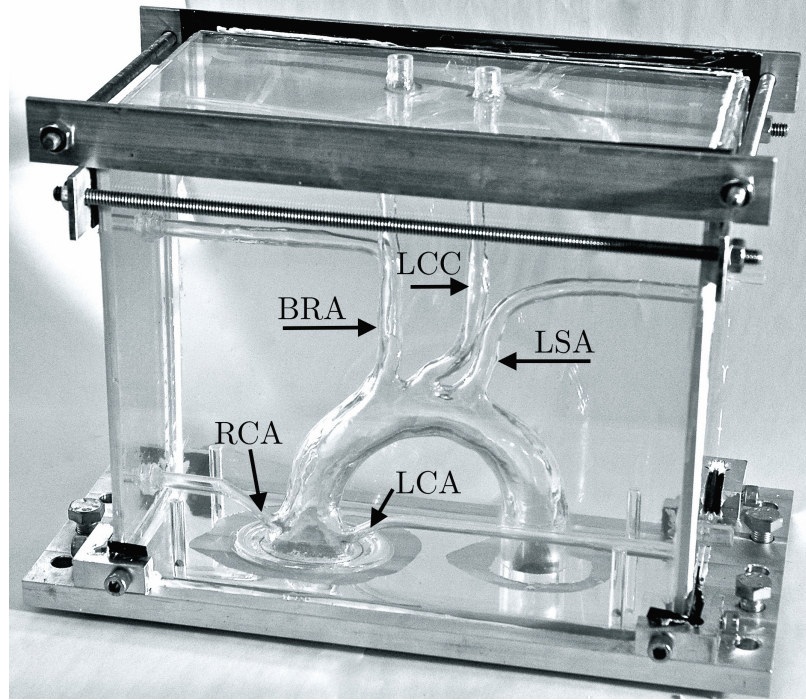


Figure 3.4: Anatomical model of aorta: RCA - right coronary artery; LCA - left coronary artery; BRA - brachiocephalic artery; LCC - left common carotid artery; LSA - left subclavian artery.

Whole human blood is a non-Newtonian fluid. Its viscosity increases at low shear strain rates due to aggregates of red blood cells forming. High levels of shear strain rate cause the aggregates to break up, effectively lowering the viscosity. Figure 3.5 shows the relationship of blood viscosity with shear strain rate [66]. At shear strain rates exceeding approximately 100 s^{-1} , the viscosity is insensitive to shear strain rate, and blood can be approximated as a Newtonian fluid [57]. In large blood vessels (greater than about 1 mm in diameter [77]), the shear strain rate normally exceeds this value; for example, White *et al.* [81] measured the peak wall shear rate in the abdominal aorta of dogs at peak systole to be between 1100 s^{-1} and 1400 s^{-1} . They found that the minimum shear strain rate occurred at end-diastole and was approximately 50 s^{-1} . For the present study in a full scale aorta model and similarly sized axisymmetric model (diameter $\gg 1 \text{ mm}$), the flow phases of most interest are during the systolic period, when shear strain rates are high. For these reasons, we determined that it was appropriate to use a Newtonian fluid for these experiments

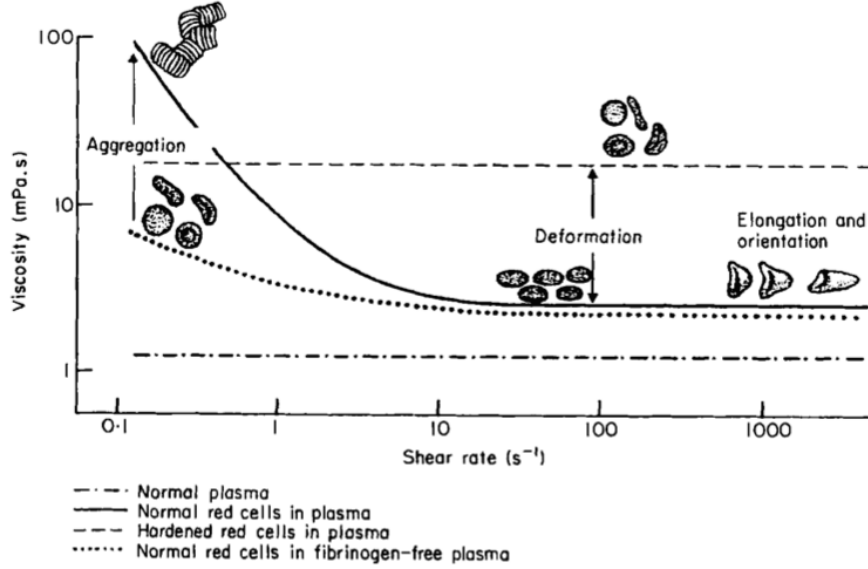


Figure 3.5: Plot of the apparent viscosity of blood *vs.* shear rate. The solid line indicates the behaviour for normal whole blood. (Adapted from Somer & Meiselman [66])

and expect that the associated error would be negligible.

The dynamic viscosity of the working fluid was measured with a dial viscometer (LVF, Brookfield) for several temperatures. The results, plotted in Figure 3.6, show that the viscosity of the fluid decreases nearly linearly with temperature. A straight line of best fit is given by:

$$\mu = -0.1042T + 6.6960 , \quad (3.1)$$

where μ is the fluid viscosity in cP, and T is the temperature in $^{\circ}\text{C}$.

The temperature of the fluid was recorded during the acquisition of all velocity measurements: the lowest recorded temperature was 20.2°C , the highest was 21.7°C , and the average was 21.2°C . For the purposes of this study, the dynamic viscosity was calculated from Equation 3.1 using the average recorded temperature, and was calculated to be 4.49 cP; this was slightly higher than the range for normal blood, which was reported by Chandran *et al.* [19] to be between 3.5 and 4.0 cP at body temperature. The largest deviation from this nominal value due to temperature variation in the lab was calculated to be 2.3%. The fluid density ρ was calculated

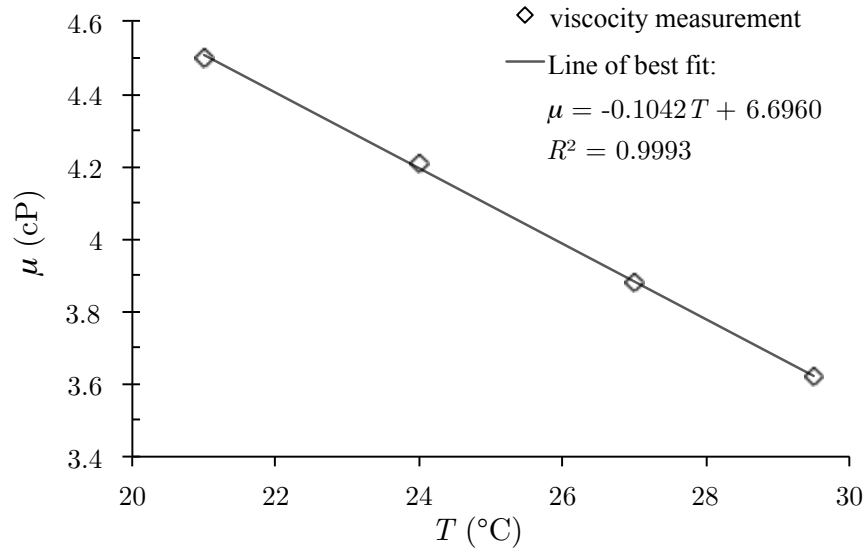


Figure 3.6: Viscosity of the working fluid measured at various temperatures.

from weighing a known volume of fluid, and was found to be 1.23 g/mL. From the dynamic viscosity and the density, the kinematic viscosity of the fluid ν was calculated to be 3.65 cSt, which was within the range of normal blood viscosity at physiological shear rates [19].

The refractive index of the fluid was fine-tuned to optimally match that of the test section walls using the following procedure:

- A sheet of paper, printed with a grid, was adhered to the back wall of the test section. If the fluid was not perfectly matched, some discontinuities of the grid lines appeared at the walls of the inner channel such as shown in Figure 3.7.
- A solution of the fluid with three-times the concentration of sodium iodide was slowly added to the loop, and the loop was run until the added fluid was thoroughly mixed.
- A photograph was taken of the grid lines, and was examined for grid line discontinuities at the channel walls.
- This procedure was repeated until no discontinuities were detectable at the channel walls (Figure 3.7).

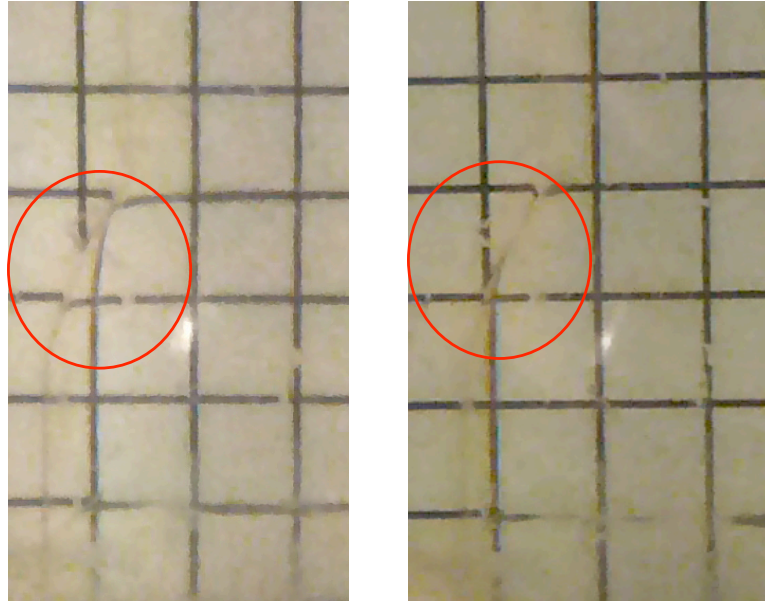


Figure 3.7: Axisymmetric section with grid for optimally matching refractive index of fluid. Left: not optimally matched, grid-line discontinuity indicated by red circle; Right: optimally matched, grid-line discontinuity minimized.

When optimally matched to the test section, the refractive index of the fluid was measured to be 1.4110 with an ABBE-3L refractometer (Bausch & Lomb, Rochester, New York).

The flow was seeded with hollow glass beads (Sphericel® 110P8; Potters Industries LLC, Malvern, PA, USA) with mean particle diameter $11.7 \mu\text{m}$, and specific gravity 1.10 ± 0.05 .

The valve used in this study was a 21 mm St. Jude Medical™(SJM) Regent™ valve. Its outer diameter was $D = 22.6 \text{ mm}$ and its inner diameter was $D_i = 19.5 \text{ mm}$ (Figure 3.3). This model is one of the most popular models in use today and has been studied extensively. It is designed to be implanted in the supra-annular position, above the aortic annulus at the base of the sinus of Valsalva (Figure 3.8). This feature allows a larger sized valve to be used for a given aortic annulus diameter than for intra-annular valves, increasing the valve effective orifice area, and reducing flow restriction.

After the model of the aorta was obtained, the BMHV was sized to appropriately

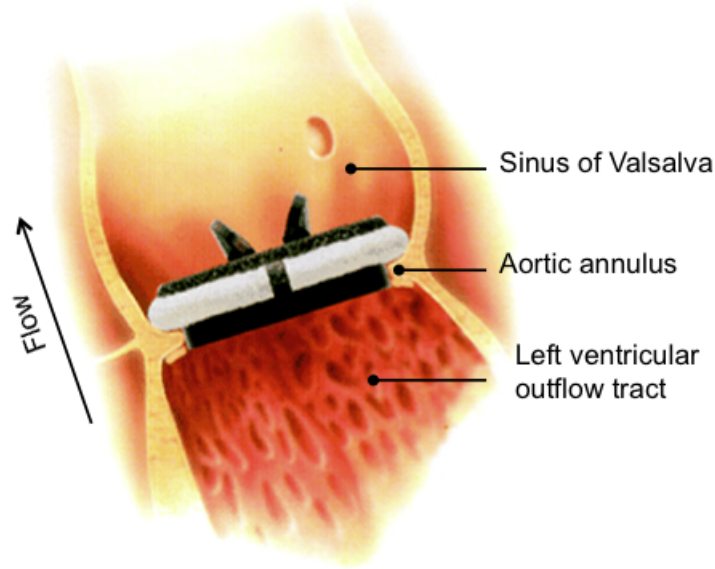


Figure 3.8: Bileaflet valve implanted in the supra-annular position. (Adapted from [76])

match the size of the aorta. A detailed explanation of how this was done is given in Appendix B.

3.1.4 Sinus remodelling

To correct for the collapse of the sinuses in the model of the aorta, and the absence of an annulus, modifications were made to the aortic root. It was determined that the sinus diameter at a location of 6 mm from the inlet of the model was an appropriate size for the diameter at the aorta's sinotubular junction (STJ). From this point and towards the model entrance, the sinuses were reduced down to a nearly circular cross-section with a diameter of approximately 21 mm, representing the aortic annulus. This modification is illustrated in Figure 3.9, where the section in purple is an insert that was fabricated and mated with the original root. The procedure used to fabricate the modifying insert is detailed in Appendix C.

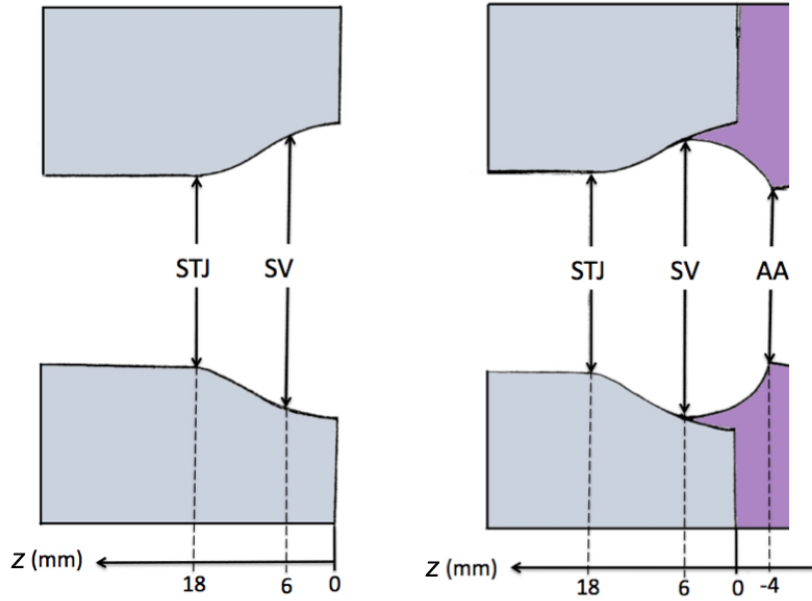


Figure 3.9: Left: sketch of the unmodified root of the aorta model with collapsed sinus. Right: sketch of the modifications to construct an annulus (in purple). STJ = sinotubular junction; SV = sinuses of Valsalva; AA = aortic annulus.

3.2 Instrumentation

3.2.1 Laser Doppler velocimetry system

A two-component laser Doppler velocimeter (LDV) system was used for measuring the local flow velocity (FiberFlow; Dantec Dynamics, Skovlunde, Denmark). The beam emitted from a 5 W argon-ion laser (model 95L-5; Lexel Laser, Freemont, CA, USA) was directed through a transmitter (FibreFlow, model 60X41; Dantec) where the beam was split in two and a Bragg cell applied a frequency shift of 40 MHz to one of the beams. A dispersion prism then separated the light of the two beams into its individual components, and only light having wavelengths of 488.0 nm (blue light) and 514.5 nm (green light) was let exit the transmitter.

The resulting four beams (two blue and two green) were then transmitted through a fibre optic cable to a probe that was mounted facing the test section. At the probe exit, the four beams, initially spaced 38 mm apart, were focused by a lens with a focal length of 160 mm to an intersection volume within the test section. The diameter

wavelength (streamwise/transverse)	514.5 nm (green) / 488.0 nm (blue)
focused beam diameter	1.35 mm
lens focal length	160 mm
initial beam spacing	38 mm
beam half angle	6.772°
frequency shift	40 MHz
fringe spacing (streamwise/transverse)	2.182 μm / 2.069 μm

Table 3.1: LDV parameters.

of each beam at its waist was 1.35 mm. The intersection of the beams formed a measurement volume that was ellipsoidal in shape with dimensions 0.07818, 0.07764, and 0.6584 mm. The LDV parameters are summarized in Table 3.1.

Light backscattered off the seeding particles was received by the probe, and transmitted to a distribution box, where two optical filters separated the colours before the light was detected by two photomultiplier tubes. These signals were then sent to a processor (BSAF50; Dantec), which sent the output to the computer for recording and analysis.

Data acquisition was controlled using Dantec BSA flow software. The voltage line that triggers the solenoid valve of the MCL to supply compressed air to the ventricle bulbs was also connected to the processor, and the software’s “cyclical add-on” option was selected. This permitted the synchronization of measurements with the phase of each pulse cycle.

The laser probe was mounted on a traversing system (Figure 3.10) that permitted its controlled displacement in the x -, y -, and z -directions. The probe position and orientation were adjusted with respect to the test section, according to the protocol described in Appendix A.2.

3.2.2 Particle image velocimetry system

A particle image velocimetry system was used to measure two-dimensional velocity fields in the vicinity of the bileaflet valve. To do this, a plane within the flow was

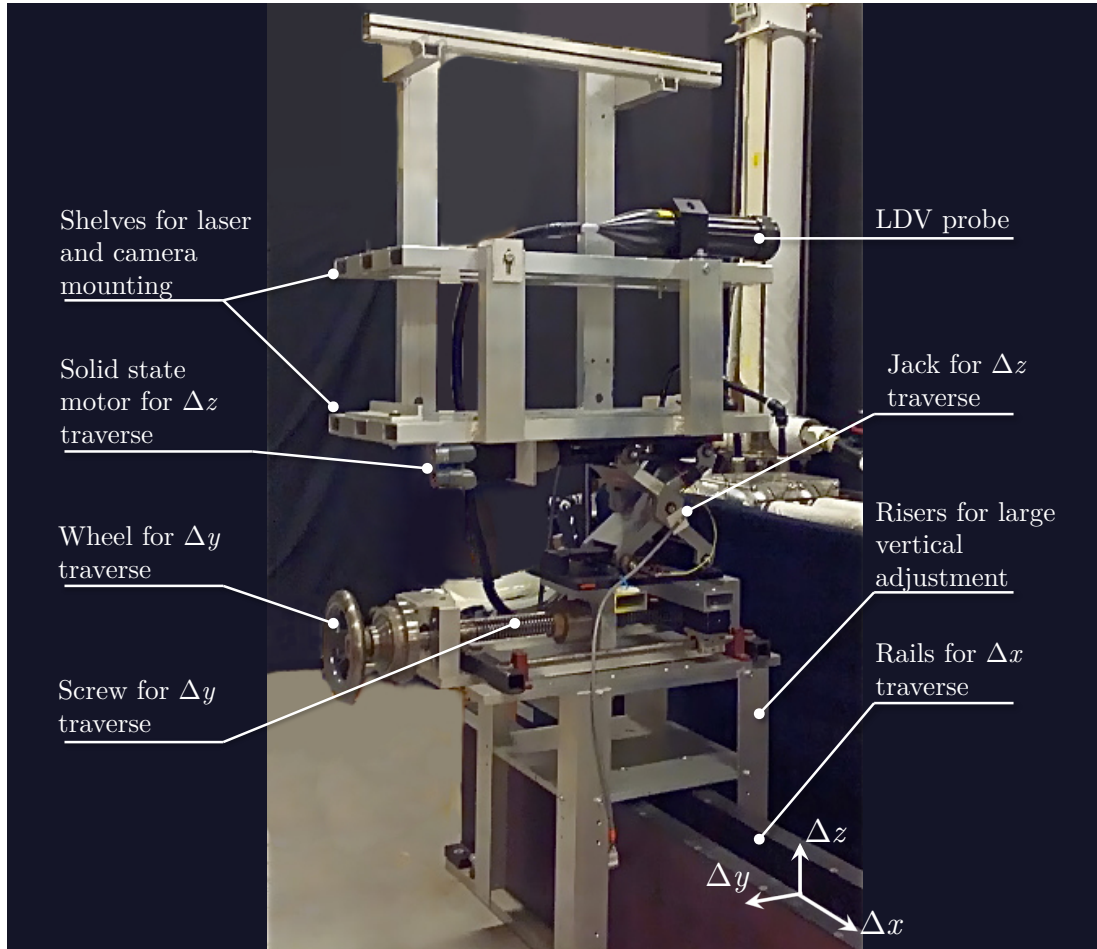


Figure 3.10: Laser and camera mounting and traversing system.

illuminated by a laser sheet. Two images were taken, one shortly after the other, of particles in this plane. The velocities of the particles were calculated from their displacements during the time interval between the images.

The laser sheet was generated by a dual cavity Nd:YAG laser (120 Solo XT; ESI, NWR Division, Sunnyvale, CA, USA) which emitted a double-pulsed beam having a wavelength of 532 nm. The power and duration of each pulse were 120 mJ and 10 ns, respectively. The beam passed through a sheet optics configuration consisting of two cylindrical lenses that transformed the beam into a thin sheet. The sheet thickness was adjusted to approximately 1.5 mm by turning the focus ring of the sheet optics, which changed the separation between the two spherical lenses. The sheet width was determined by the focal length of the cylindrical lens. For the measurements in this

work, a cylindrical lens having a focal length of -20 mm was used.

The laser head, from which the light sheet was emitted, was mounted on the laser traversing system described in Section 3.2.1. The sheet was positioned and oriented to illuminate the desired plane within the test section by maneuvering the traverse, and by positioning a mirror that was mounted in the path of the beam to change the sheet's direction. Details of the procedure used to align the laser sheet to the desired plane are described in Appendix A.3. Two configurations of the equipment were used for illuminating the front and side planes of the test section, and are illustrated in Figure 3.11.

A CCD camera (ImagerProX 4M; LaVision, Göttingen, Germany) was mounted on the traversing system and focused onto the illuminated plane. The resolution of the camera was 2048×2048 pixels. The camera was interfaced with a macroscopic lens (Tamron 90 mm), and an optical filter having a narrow waveband (10 nm) centred about 532 nm to filter out ambient light.

The action of the laser and camera were synchronized with the voltage signal of the MCL's solenoid valve via a programmable timing unit (PTU9; LaVision). This permitted phase locked measurements, and subsequent phase averaging. All of the components of the PIV system were connected to a computer, through which the data acquisition was controlled using image capture and analysis software (DaVis 8.1; LaVision).

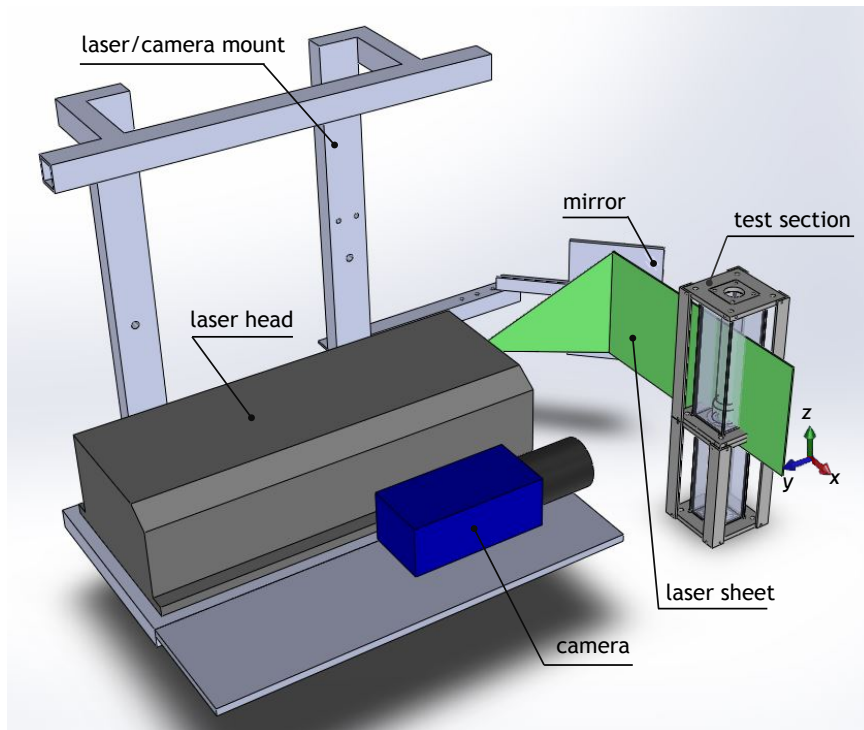
The PIV system was calibrated according to the protocol outlined in Appendix A.3. The PIV recording parameters are summarized in Table 3.2

3.3 Post-Processing Procedures

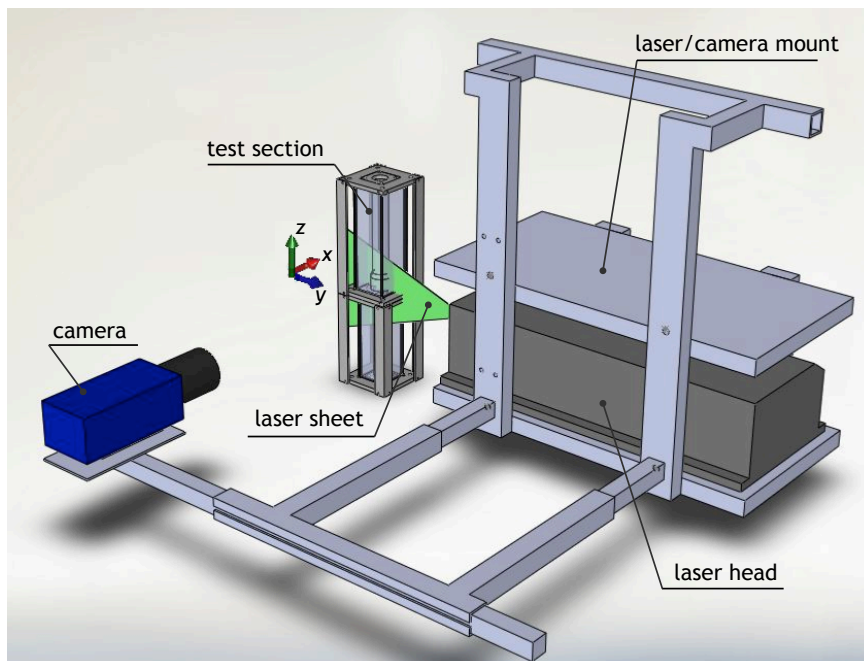
3.3.1 Single-point velocity measurements

Phase sorting

The velocity measurements acquired by LDV at each location were sorted into bins, according to their arrival time with respect to the start of the parent cycle, which was marked by the opening of the solenoid air valve to the ventricle bulbs of the MCL.



(a)



(b)

Figure 3.11: Equipment configurations for PIV measurements in the x - z (front) plane (a), and the y - z (side) plane (b). (The coordinate axes shown here are not positioned at the origin for the sake of visual clarity).

laser sheet wavelength	532 nm
sheet thickness	1 – 1.5 mm
laser lens focal length	-20 mm
camera lens focal length	90 mm
camera lens aperture f#	8 or 11
optical filter	10 nm bandwidth, centred about 532 nm
image size	2048 × 2048 pixels
pulse delay	70 – 1000 μ s
typical particle displacement	2 – 8 pixels

Table 3.2: PIV experimental parameters.

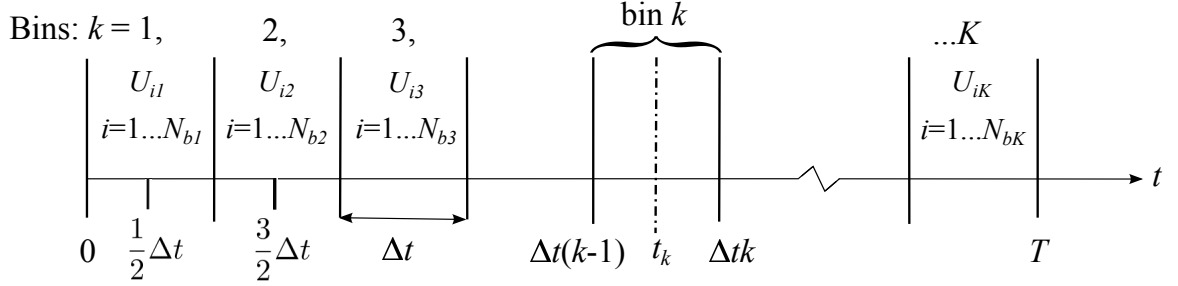


Figure 3.12: Sketch representing the bins for phase averaging the LDV measurements.

As illustrated in Figure 3.12, the cycle period T was discretized into K bins, where the width of each bin Δt is

$$\Delta t = \frac{T}{K}. \quad (3.2)$$

A velocity measurement U_i would be allocated to the bin which includes the measurement's arrival time t_i within its range. The range of the k^{th} bin is given by

$$\Delta t(k-1) \leq t < \Delta tk. \quad (3.3)$$

The reference time t_k of each bin is located at the bin centre, and is given by

$$t_k = \Delta t \left(k - \frac{1}{2} \right), \quad k = 1, 2, \dots, K. \quad (3.4)$$

The x -direction velocity measurements in bin k are denoted as U_{ik} , $i = 1, 2, \dots, N_{bk}$, where N_{bk} is the number of measurements in the bin. The velocity measurements

within a single cycle m are denoted as U_{im} . In this parameter, $i = 1, 2, \dots, N_{cm}$, where N_{cm} is the number of measurements acquired for cycle m , and $m = 1, 2, \dots, M$, where M is the total number of cycles acquired. The total number of measurements acquired at a location can be determined as the sum of the number of measurements over all cycles, or alternatively as the sum of the number of measurements over all bins, i.e. as $N = \sum_{m=1}^M N_{cm} = \sum_{k=1}^K N_{bk}$.

The **phase-averaged velocity**, *i.e.* the averaged velocity for each bin, was calculated as

$$\langle U \rangle_k = \frac{1}{N_{bk}} \sum_{i=1}^{N_{bk}} U_{ik}, \quad k = 1, 2, \dots, K. \quad (3.5)$$

The fluctuations within each bin k were determined by subtracting the bin's mean from each measurement as

$$u_{ik} = U_{ik} - \langle U \rangle_k, \quad i = 1, 2, \dots, N_{bk}. \quad (3.6)$$

From these, the standard deviation (r.m.s.) of fluctuations for each bin was then calculated as

$$u'_k = \sqrt{\frac{1}{N_{bk} - 1} \sum_{k=1}^{N_{bk}} u_{ik}^2}, \quad k = 1, 2, \dots, K. \quad (3.7)$$

The calculations described here were performed by the LDV software (BSA Flow Software; Dantec). They were performed similarly for the transverse velocity V and streamwise velocity W , resulting in values of $\langle V \rangle_k$, v_{ik} , and v'_k , and $\langle W \rangle_k$, w_{ik} , and w'_k , respectively. These velocity components are illustrated with the coordinate system and BMHV in Figure 3.13.

Additional data manipulation

Next, the measurements were scanned to identify outlier measurements, which were discarded before further data processing. The details of this procedure are provided in Appendix D. Following the removal of outliers, means and standard deviations $\langle U \rangle_k$ and u'_k were re-calculated according to Equations 3.5 and 3.7, respectively. To accomplish this, the data were first re-sorted into the phase bins described above. The measurements were also sorted by cycle into an array having three dimensions:

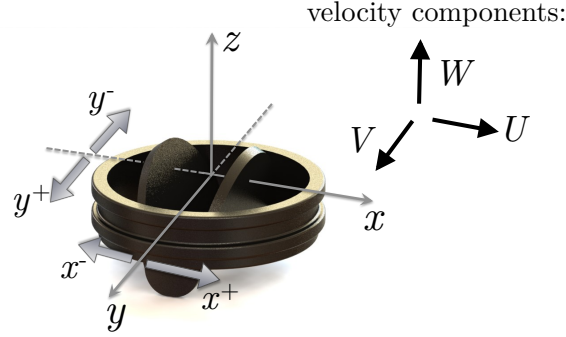


Figure 3.13: Sketch of velocity components with coordinate system and BMHV.

measured values, variables, and cycles. This format facilitated subsequent computations.

To avoid excessively large LDV files, which are cumbersome for the LDV software and would impede its performance, the sinus region measurements were acquired in two separate data sets. These two sets were joined into a single array for each position, before being processed.

All of these manipulations and calculations were performed by using in-house MATLAB programs, and were done for the streamwise and transverse velocity components.

Flow rate calculation

The cardiac output (a.k.a. the bulk flow rate) was calculated as the time average of the time dependent volume flow rate through the test section:

$$\overline{Q} = \lim_{m \rightarrow \infty} \frac{1}{m} T \int_0^{mT} Q(t) dt , \quad (3.8)$$

where the overbar indicates the time-averaging operation. For axisymmetric flow in a tube, the instantaneous flow rate is given by

$$Q(t) = 2\pi \int_0^R W(r, t) r dr , \quad (3.9)$$

where $W(r, t)$ is the axial velocity at a radial position r and time t , and R is the tube radius. It then follows that the cardiac output is given by

$$\overline{Q} = \overline{2\pi \int_0^R W(t)r dr} \quad (3.10a)$$

$$= 2\pi \int_0^R \overline{W}r dr, \quad (3.10b)$$

where $\overline{W}$ is the time averaged velocity at the radial position r .

Two methods were used to approximate the cardiac output from the LDV measurements. In the first method, the time averaged velocity was calculated for each position, before being integrated in space, according to Equation 3.10b. In the second method, the phase-averaged velocity $\langle W \rangle_k$ was integrated in space, prior to averaging in time. The first method offers a better estimate of the time averaged flow rate, as it is not subject to binning error, but provides no information of the flow rate distribution in time. The second method, although subject to binning error, provides an estimate of the bulk flow rate at each time bin. The detailed calculations of the two methods are described in the following.

Method I

The time averaged velocity for every cycle was calculated as

$$\overline{W}_m = \frac{1}{T} \left[\left(\sum_{i=1}^{N_{cm}-1} W_{im}(t_{(i+1)m} - t_{im}) \right) + W_{1m}(t_{1m}) + W_{N_{cm}m}(T - t_{N_{cm}m}) \right], \quad (3.11)$$

where t_{im} is the arrival time of measurement W_{im} , $t_{(i+1)m}$ is the arrival time of the subsequent measurement, and $t_{N_{cm}m}$ is the arrival time of the last measurement of the cycle. The time averaged velocity at a position was then calculated by averaging $\overline{W}_m$ over all the cycles as

$$\overline{W} = \frac{1}{M} \sum_{m=1}^M \overline{W}_m. \quad (3.12)$$

To estimate the bulk flow rate $\overline{Q}$, this result was then integrated over the tube radius, according to Equation 3.10b, using the trapezoidal integration method. A no-slip condition was assumed at the walls, and $\overline{W}(x = R) = 0$ was enforced. For each axial location z , four independent radii were available to be used for the calculation. These

were the positive and negative halves of the x and y profiles, denoted as x^+ , x^- , y^+ , and y^- . For example, the average flow rate calculated from the measurements of the x^+ radii was found as

$$\overline{Q}_{x^+} = 2\pi \int_0^R \overline{W(x^+)} x dx. \quad (3.13)$$

Similarly, $\overline{Q}_{x^-}$, $\overline{Q}_{y^+}$, and $\overline{Q}_{y^-}$ were obtained. These results were then averaged to provide a representative cardiac output $\overline{Q}$, as follows:

$$\overline{Q}_x = \frac{1}{2}(\overline{Q}_{x^+} + \overline{Q}_{x^-}) \quad (3.14a)$$

$$\overline{Q}_y = \frac{1}{2}(\overline{Q}_{y^+} + \overline{Q}_{y^-}) \quad (3.14b)$$

$$\overline{Q} = \frac{1}{2}(\overline{Q}_x + \overline{Q}_y). \quad (3.14c)$$

Method II

Here, the phase averaged velocities $\langle W \rangle_k$ were integrated across the tube radius, for each time bin, to obtain the phase averaged bulk flow rate as

$$\langle Q \rangle_k = 2\pi \int_0^R \langle W \rangle_k x dx. \quad (3.15)$$

The result was then integrated over the cycle time to obtain the second estimate of cardiac output as

$$\overline{Q}^* = \frac{1}{T} \int_0^T \langle Q \rangle_k dt, \quad dt = k\Delta t. \quad (3.16)$$

As in Method I, the four available radii were used to obtain $\overline{Q}_{x^+}^*$, $\overline{Q}_{x^-}^*$, $\overline{Q}_{y^+}^*$, and $\overline{Q}_{y^-}^*$, which were then averaged to obtain $\overline{Q}^*$, in a manner similar to that described by Equation (3.14).

In both Method I and Method II, the calculation of $\overline{Q}$ is based on the assumption of axisymmetric flow. This assumption is fairly accurate for measurements taken far downstream or far upstream of the valve. It is, however, significantly inaccurate for the measurements taken in the sinus region, immediately downstream of the valve, because the bileaflet valve is not axisymmetric and its presence and operation disturb the flow in a non-axisymmetric way. Ideally, for perfectly axisymmetric flow, $\overline{Q}_{x^+}$

should equal $\overline{Q}_{x-}$, $\overline{Q}_{y+}$ should equal $\overline{Q}_{y-}$, and $\overline{Q}_x$ should equal $\overline{Q}_y$. For flow in the wake of the BMHV, which is symmetric about its x -axis and about its y -axis, but is not axisymmetric, significant differences between the calculated values of $\overline{Q}_x$ and $\overline{Q}_y$ are expected. Therefore, the estimates of $\overline{Q}$ in these regions would have significant uncertainties.

3.3.2 Planar velocity measurements

Image pre-processing

The images acquired with PIV were processed using the image capture and analysis software (Davis 8.1.4). For each camera view, an averaged image was obtained from a set of 1000 raw particle images. The result was such that the particle images blended to form a homogeneous background, while stationary features in the field of view (such as physical features of the apparatus and reflections) remained visible (Figure 3.14b). The averaged image was subtracted from each raw image prior to analysis, effectively removing the stationary components, leaving only the particle images. This procedure significantly reduced the number of spurious vectors that would otherwise have resulted from the stationary components. Samples of the raw, averaged, and resulting difference particle images for the axisymmetric model are shown in Figure 3.14.

The particle intensities were then normalised by the minimum and maximum intensities of the surrounding pixels within a 5×5 pixel window. This homogenized the intensities, so that particles would equally contribute to the correlation, regardless of size. A polygon that outlined the channel geometry was applied as a mask, such that only pixels inside the geometry were processed. Additionally, a lower and an upper threshold were applied, setting pixel intensities less than 75 or greater than 2500 to zero. Then, pixels with intensities equal to zero were eliminated by setting their value to one.

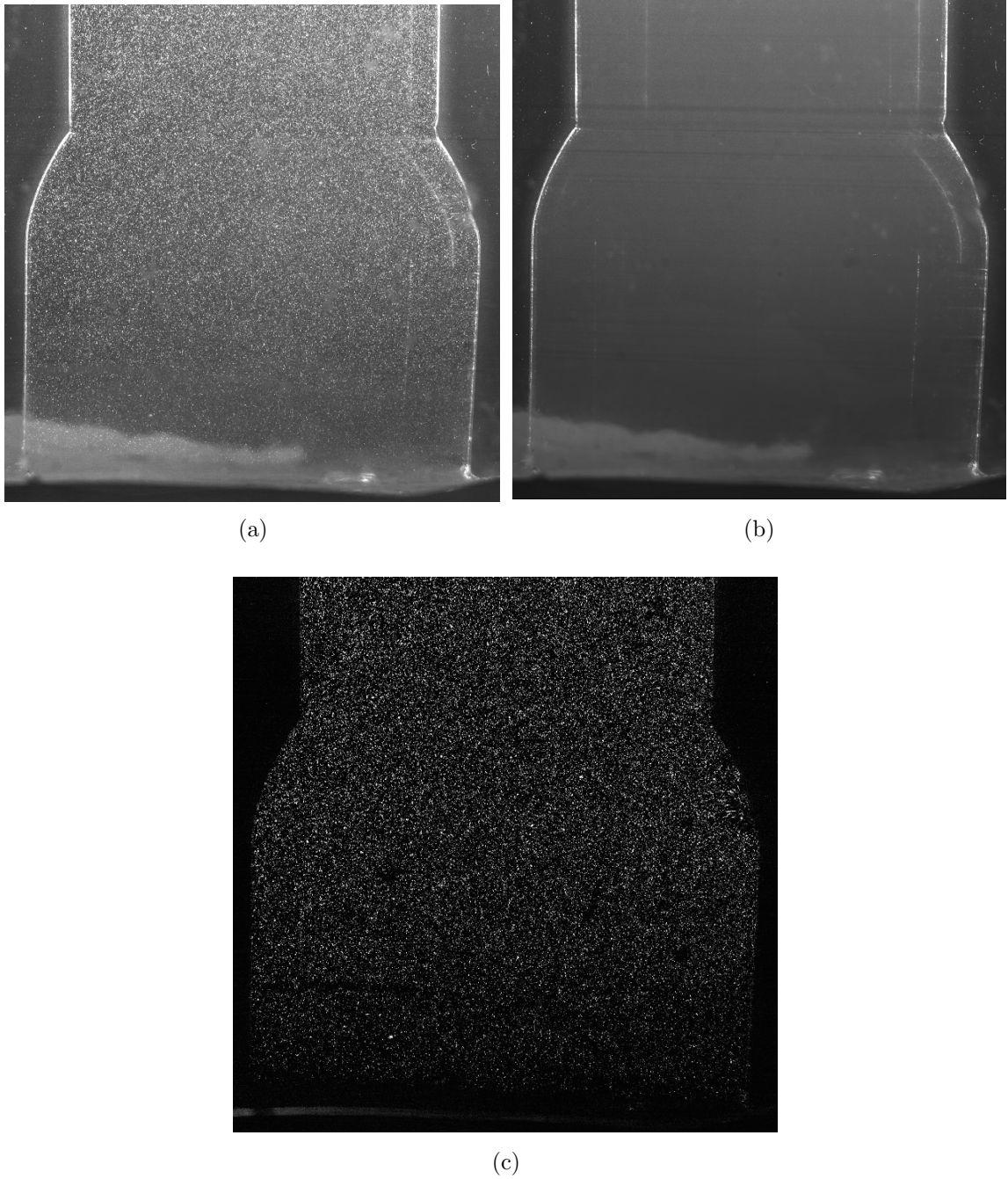


Figure 3.14: Sample particle images for the axisymmetric model: (a) raw particle image; (b) average of 1000 particle images; (c) particle image after removal of average.

Vector calculation

Velocity vectors were then calculated using cross correlation algorithms between the image pairs. Multiple iterative passes were made, using successively smaller sized interrogation windows. For the measurements made in the axisymmetric test section, the interrogation window size was initially 48×48 pixels and was decreased to 12×12 pixels over four subsequent passes. The anatomical model contained some imperfections such as inhomogeneities in the silicone, which caused some streaks to appear in the calculated velocity field. To minimize these effects, a larger interrogation window size was used for the measurements in the anatomical aorta model: the initial window size was 64×64 pixels, and the final window size was 24×24 pixels. For all passes in both models, a 50% overlap between neighbouring interrogation windows was used. The software’s “high accuracy mode” was used for final passes. The parameters used for processing the PIV measurements are summarized in Table 3.3.

Figure 3.15 shows zoomed-in portions of particle images acquired from the axisymmetric and anatomical models with squares showing the respective final interrogation window sizes. Adrian & Westerweel [1] recommended the use of 5 to 15 particle images per interrogation area, and suggested that the optimal value was 10; Figure 3.15 shows that in typical windows of the axisymmetric and anatomical models the particle densities were approximately 8 and 12, which are within the recommended range.

The result was a set of M velocity fields, each containing either 341×341 or

test section:	axisymmetric	anatomical
initial window size	48×48 pixels	64×64 pixels
number of refinement steps	4	2
final window size	12×12 pixels	24×24 pixels
vector grid size	341×341	170×170
filter	Gaussian smoothing (3×3 window)	

Table 3.3: PIV processing parameters.

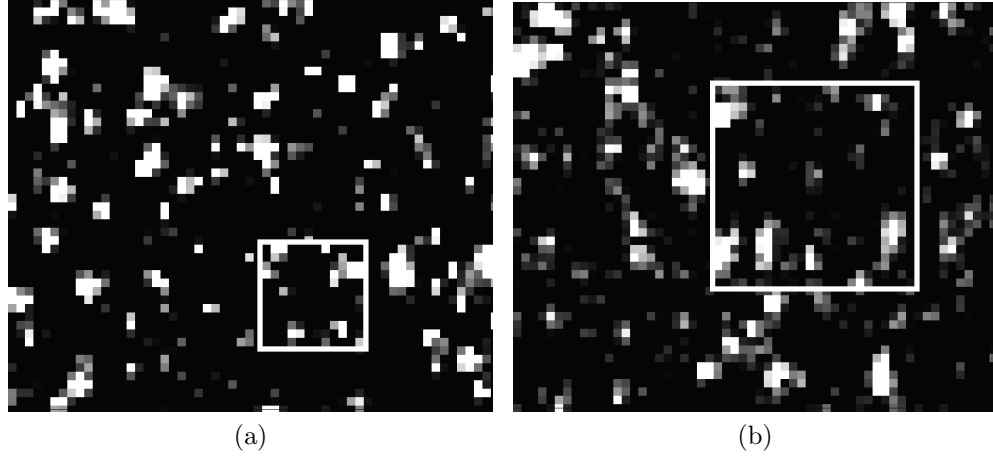


Figure 3.15: Particle images with interrogation windows for axisymmetric model (left) and anatomical aorta model (right).

170×170 two-component velocity vectors for the axisymmetric model or the anatomical model, respectively. The vector fields were then post-processed with a median filter which removed vectors that differed more than three standard deviations from the average of their neighbours. Groups containing less than three vectors were also removed. Any empty spaces were assigned a vector from an interpolation between the neighbouring vectors. Seeding and laser light conditions were adjusted so that there were very few empty spaces ($< 0.1\%$). In order to reduce noise related error in the subsequent calculation of viscous shear stress, the vector fields were further smoothed with a Gaussian smoothing filter (using a 3×3 window). This treatment was found by Luff *et al.* [54] to be able to reduce the average uncertainty of second order calculations from PIV data from $\pm 37\%$ to $\pm 4\%$. The spatial coordinates were then shifted to position the origin at the mid-line of the bileaflet valve, at the downstream edge of the valve housing.

Most generally, the velocities for the node located at the i^{th} column and the j^{th} row of the m^{th} velocity field measured for a phase are denoted U_{ijm} , V_{ijm} , and W_{ijm} in the x , y , and z directions, respectively. All of the following calculations are applied to each node in every velocity field, so, for simplicity, the i and j subscripts will be

dropped. The planar velocity magnitudes were defined at each node (i,j) as

$$q_{xz} = \sqrt{U^2 + W^2}, \quad (3.17a)$$

$$\text{and } q_{yz} = \sqrt{V^2 + W^2}, \quad (3.17b)$$

for the x - z and the y - z planes, respectively. Vector statistics were calculated from each ensemble of velocity maps. The average velocities were determined at each node as

$$\langle U \rangle = \frac{1}{M} \sum_{m=1}^M U_m, \quad (3.18)$$

and similarly for $\langle V \rangle$, $\langle W \rangle$, $\langle q_{xz} \rangle$, and $\langle q_{yz} \rangle$. The r.m.s. velocity fluctuations were calculated for each node as

$$u' = \sqrt{\frac{1}{M-1} \sum_{m=1}^M (U_m - \langle U \rangle)^2}, \quad (3.19)$$

and similarly for v' and w' .

Viscous shear stress calculation

The two-dimensional viscous shear stress tensor τ is given by

$$\tau = \begin{vmatrix} \tau_{xx} & \tau_{xy} \\ \tau_{yx} & \tau_{yy} \end{vmatrix} \quad (3.20a)$$

$$= 2\mu \begin{vmatrix} E_{xx} & \frac{1}{2}(E_{xy} + E_{yx}) \\ \frac{1}{2}(E_{xy} + E_{yx}) & E_{yy} \end{vmatrix}, \quad (3.20b)$$

where the strain rate components E_{xx} , E_{xy} , E_{yx} , and E_{yy} were calculated from the velocity gradients, using a second order central differencing scheme, as

$$E_{xx} = \frac{\partial U}{\partial x} \approx \frac{U(i+1, j) - U(i-1, j)}{2\Delta x}, \quad (3.21a)$$

$$E_{xy} = \frac{\partial U}{\partial y} \approx \frac{U(i, j+1) - U(i, j-1)}{2\Delta y}, \quad (3.21b)$$

$$E_{yx} = \frac{\partial V}{\partial x} \approx \frac{V(i+1, j) - V(i-1, j)}{2\Delta x}, \quad (3.21c)$$

$$\text{and } E_{yy} = \frac{\partial V}{\partial y} \approx \frac{V(i, j+1) - V(i, j-1)}{2\Delta y}. \quad (3.21d)$$

A coordinate invariant maximum shear stress τ_{max} was defined as

$$\tau_{max} = \frac{1}{2} (\sigma_{max} - \sigma_{min}), \quad (3.22)$$

where σ_{max} and σ_{min} are the maximum and minimum principal shear stresses of τ . This was equivalently calculated as

$$\tau_{max} = 2\mu \sqrt{\frac{(E_{xx} - E_{yy})^2}{4} + \frac{(E_{xy} - E_{yx})^2}{4}}. \quad (3.23)$$

The viscous shear stress measurement was highly sensitive to the grid resolution of the PIV measurements. A grid refinement test was performed and is described in Appendix E.1. From this, it was reasoned that the viscous shear stress calculated for the anatomical measurements was underestimated as a result of the lower vector resolution. To correct for this, a corrective mapping algorithm was derived from the results in the axisymmetric model and applied to the viscous shear stress measurements in the anatomical model. The derivation of the corrective mapping algorithm is detailed in Appendix E.2.

Turbulent stress calculation

As a measure of the turbulence in the wake of the BMHV, turbulent stresses were calculated from the velocity fluctuations; the Reynolds stress tensor for the x - z plane

is given, for example, as

$$\mathbf{RS} = \rho \begin{vmatrix} \langle u^2 \rangle & \langle uw \rangle \\ \langle uw \rangle & \langle w^2 \rangle \end{vmatrix}. \quad (3.24)$$

A 2D coordinate invariant maximum turbulent stress RS_{max} was calculated from the principal stresses of the Reynolds stress tensor in the same way as was done for the viscous shear stress [35].

Coherent structure detection

Coherent vortical structures past the valve were identified by the swirling strength criterion λ_{ci} . This criterion proposed by Zou *et al.* [91] identifies vortex structures from the imaginary part of the complex eigenvalue of the velocity gradient tensor $\mathbf{E}$ [59]. This parameter is called the swirling strength and is denoted as λ_{ci} :

$$\lambda_{ci} = \text{Im}[\text{eig}(\mathbf{E})], \quad (3.25a)$$

$$= \frac{1}{2} \text{Im} \left[\sqrt{(E_{xx} - E_{yy})^2 + 4E_{xy}E_{yx}} \right], \quad (3.25b)$$

where the velocity gradient tensor $\mathbf{E}$ is

$$\mathbf{E} = \begin{vmatrix} E_{xx} & E_{xy} \\ E_{yx} & E_{yy} \end{vmatrix}. \quad (3.26a)$$

The swirling strength was calculated from instantaneous PIV measurements, using the PIV software (Davis 8.1).

3.4 Uncertainty Estimates

3.4.1 The positioning system

The positioning system was mounted on rails that were aligned with the x -axis and was moved manually along them to the desired position, which was indicated by a dial gauge with precision 0.03 mm (0.001 in). The manual positioning of the traverse in this direction was repeatable to within 0.1 mm (0.005 in). The system was moved in the y -direction by turning a wheel that rotated a lead screw, which was also precise

to 0.03 mm (0.001 in), and positioning was repeatable to within 0.05 mm (0.002 in). The system moved in the z -direction on a jack controlled by a solid state motor, with an angular precision of 0.2 degrees of rotation, which translated to a precision of 0.0001 mm of vertical displacement; however, the vertical translation was calibrated with gauges precise to 0.03 mm. Alignments in this direction were made by visually aligning a laser beam to measured markers on the test section, and this positioning was repeatable to within 0.5 mm.

3.4.2 Dynamic response of seeding particles

For both LDV and PIV measurements, the motions of tracer particles were measured to estimate the velocities of the fluid surrounding them. The particles, however, were subject to drag, buoyancy and gravitational forces that may have caused the particle velocity u_P to differ slightly from the fluid velocity u_F .

According to Tavoularis [70], the motion of a spherical particle within a fluid can be approximated as a first order system, having a time constant

$$\tau_P = \frac{\rho_P d_P^2}{18\mu}, \quad (3.27)$$

where d_P is the particle diameter and ρ_P is the particle density. For a particle with $d_P = 11.7 \mu\text{m}$, and $\rho_P = 1.1 \text{ g/mL}$, the time constant was $1.9 \mu\text{s}$.

The particles used in this study were less dense than the working fluid. Following a sufficiently long time ($t > 6\tau_P$), a particle would have reached a terminal rise velocity

$$(u_P - u_F)_\infty = (1 - \rho/\rho_P)g\tau_P. \quad (3.28)$$

For the current work, the terminal rise velocity was determined to be $2.2 \mu\text{m/s}$, which is entirely negligible.

3.4.3 Laser Doppler velocimetry

The only source of significant uncertainty in the LDV measurements was from velocity bias. This is the bias of the mean velocity measurement towards higher velocities, as a result of a greater proportion of higher speed particles crossing the measurement

volume than slower ones. An estimate of this bias is given by Tavoularis [70] as

$$\frac{\overline{U}_m}{\overline{U}} \approx 1 + \frac{\overline{u^2}}{\overline{U}^2}, \quad (3.29)$$

where $\overline{U}_m$ is the mean velocity measured by LDV, $\overline{U}$ is the true mean velocity, and $\overline{u^2}$ is the velocity variance. For the centre-line measurement at peak flow, where $\overline{U}$ was measured to be 1.99 m/s and $\overline{u^2}$ was calculated to be 0.35 m/s, the velocity bias ratio was 1.03, which implies that the velocities may have been over-estimated by 3%. To reduce this error, the measurements were weighted by their transit times through the measurement volume for the calculations of the velocity means and standard deviations.

3.4.4 Particle image velocimetry

The velocity vectors U from PIV measurements were calculated according to

$$U = \frac{\alpha_M \Delta X}{\delta t}, \quad (3.30)$$

where α_M is the magnification factor in mm/pixel, determined by calibration, ΔX is the particle image displacement in pixels, and δt is the time separation between images in seconds. For a typical measurement at the peak flow phase in the axisymmetric test section, nominal values of these factors were $\alpha_M = 0.017$ mm/pixel, $\Delta X = 6$ pixels, and $\delta t = 65 \mu\text{s}$, resulting in a velocity $U = 1.57$ m/s. Errors in each of these parameters contributed to errors in the velocity measurement according to

$$\frac{\delta U}{|U|} = \sqrt{\left(\frac{\delta \alpha_M}{\alpha_M}\right)^2 + \left(\frac{\delta(\Delta X)}{\Delta X}\right)^2 + \left(\frac{\delta(\delta t)}{\delta t}\right)^2} \quad (3.31)$$

where $\delta \alpha_M$, $\delta(\Delta X)$, and $\delta(\delta t)$ are the respective errors for α_M , ΔX , and δt .

Magnification factor α_M

The magnification factor α_M is the ratio between the physical length of the measurement plane in mm and the corresponding image length in pixels. A number of factors can contribute to uncertainty in its determination.

- The standard deviation of calibration σ was reported by the LaVision software to be 0.6 pixels. The calibration uncertainty of α_M , estimated as 2σ , was 0.06%.
- According to the Specialist Committee on Uncertainty Analysis of the 25th ITTC [67], image distortion from the camera lens may cause as much as 0.5% error in the calculation of the magnification factor.
- Error may have resulted from the calibration board being slightly off-set from the measurement plane. In the present study, the calibration was performed at the outside wall of the test section prior to translating the optical system to the measurement plane. The distance translated was the physical distance between the outside wall and the measurement plane $d_{pl} = 29.4 \pm 0.5$ mm divided by the optical ratio $\alpha_\eta \approx \eta$, where η is the refractive index of the fluid and is 1.41 ± 0.005 . As such, the translated distance was 20.8 ± 0.4 mm. The act of translating the camera introduced an additional uncertainty of the measurement plane location equal to 0.1 mm (0.004 in), which was effectively negligible. The resulting uncertainty in the position of the calibration board relative to the measurement plane was 0.4 mm. Using the sensitivities of the magnification factor to the camera-target distance, as described by Specialist Committee on Uncertainty Analysis of the 25th ITTC [67], the resulting uncertainty in α_M was found to be 0.2%.
- Error may also have resulted from the calibration board not being parallel to the laser light sheet. For a maximum deviation of 2° , the resulting uncertainty of α_M was 0.1%.

The combined relative uncertainty of the magnification factor $\frac{\delta\alpha_M}{\alpha_M}$ was 0.5%. This was propagated as a bias error to the velocity calculation because it was introduced in calibration and affected equally all of the measurements made for the corresponding data set.

Displacement of particle image ΔX

The determination of the particle displacement ΔX was subject to both precision and bias errors.

Temporal and spatial fluctuation in laser power could have resulted in some error in the detection of particle images. According to the Specialist Committee on Uncertainty Analysis of the 25th ITTC [67], under well controlled experimental conditions, this error can be estimated to be $\sqrt{2}d_p/10$, where d_p is the particle diameter. This results in an 2% precision uncertainty of the particle displacement.

Another precision error may have occurred during the correlation process due to the mis-matching of particle images. The Specialist Committee on Uncertainty Analysis of the 25th ITTC [67] suggests that this precision error could be 0.2 pixels, or in this case, 3% of the particle displacement. Sub-pixel analysis could have resulted in an additional 0.5% precision error in the determination of ΔX .

The combined precision uncertainty of ΔX was estimated to be 4%.

A bias error would have resulted from a misalignment of the camera view angle with the normal of the measurement plane. For a maximum mis-alignment of 5° , the measured velocity would have been under-estimated by 0.4%.

Time interval δt

The time separation of the laser pulses was controlled by the timing unit which has an uncertainty of 5 ns. The total uncertainty in δt occurring between two pulses was 10 ns. For a typical δt of 70 μs , the uncertainty was 0.01% which was negligible.

Summary of PIV uncertainty

The combined bias limit for the velocity measurement was 0.7%, and the combined precision limit was 4%, resulting in a total uncertainty of 4% for instantaneous velocity measurements. For phase-averaged measurements however, the precision uncertainty of the mean value was reduced by $1/\sqrt{N}$, where N is the number of samples in the ensemble. For the ‘key phase’ measurements, for which 1000 samples were recorded, the precision uncertainty was reduced to 0.1% and the total uncertainty became 0.7%. For the other phases, for which 250 measurements were taken, the precision uncertainty was 0.3% and the total uncertainty was 0.7%.

Chapter 4

Results: Velocity Measurements in an Axisymmetric Channel

4.1 Single-point velocity measurements

4.1.1 Measurement locations and sampling conditions

The bileaflet valve was mounted within the flange at the inlet of the axisymmetric test section, which will be referred to as the “valve housing flange”. A rectangular, left-handed coordinate system was defined such that the z -axis was normal to the plane in which the valve was seated and positive in the downstream direction (Figure 4.1). The origin was at the valve centre, at the downstream edge of the valve housing ring. The x - and y -axes ran perpendicular and parallel to the valve’s plane of symmetry, respectively. The positive and negative sides of the x -axis were denoted as x^+ and x^- , respectively, and similar notation was used for the y -axis. Distances were normalized by the outer diameter of the BMHV, $D = 22.6$ mm.

Axial velocity W , and transverse velocity U were measured at locations along the x - and y -directions, on various planes at different axial (z) positions. All axial locations were referenced from the top of the valve housing flange. Profiles were

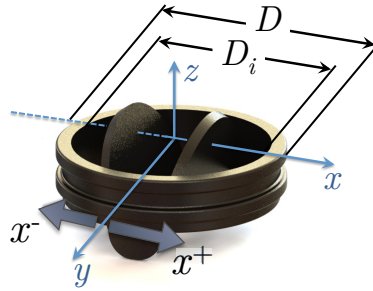


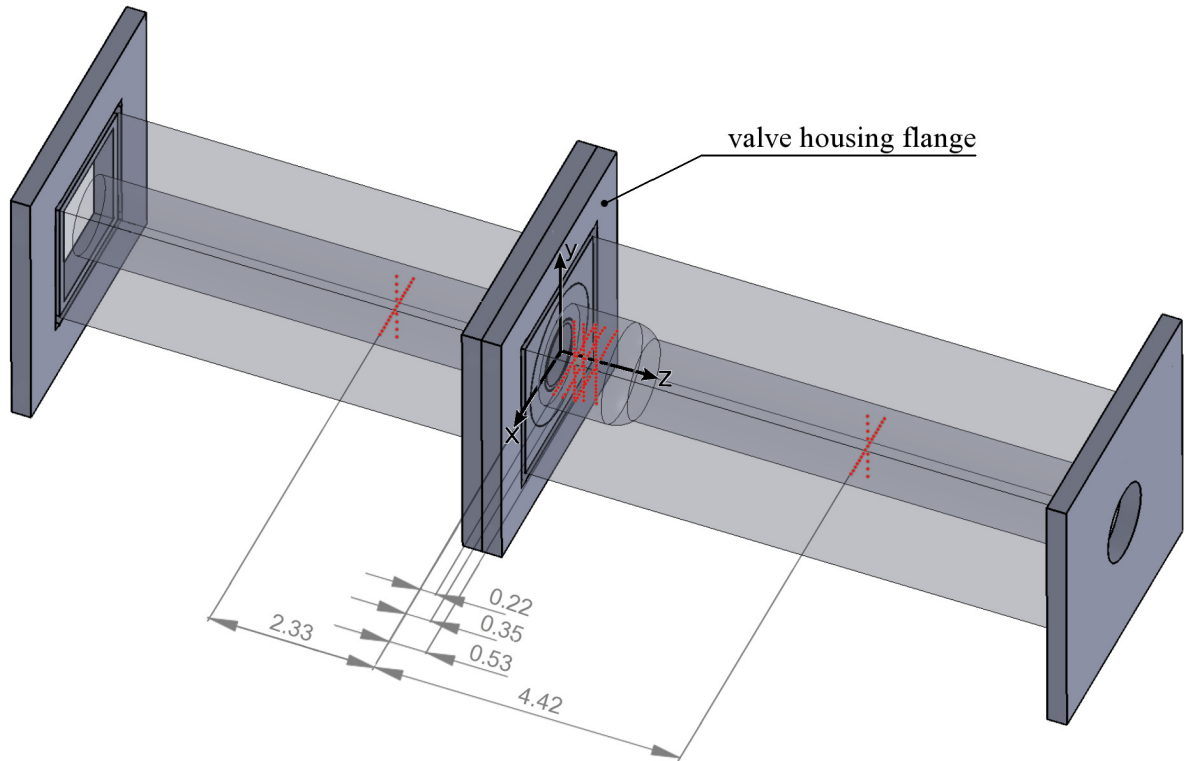
Figure 4.1: Sketch of coordinate system with BMHV.

measured upstream of the valve at $z/D = -2.33$, downstream of the valve in the axisymmetric sinus at $z/D = 0.22, 0.35$, and 0.53 , and further downstream, past the end of the sinus at $z/D = 4.42$. The measurement positions are illustrated in Figure 4.2 as red dots.

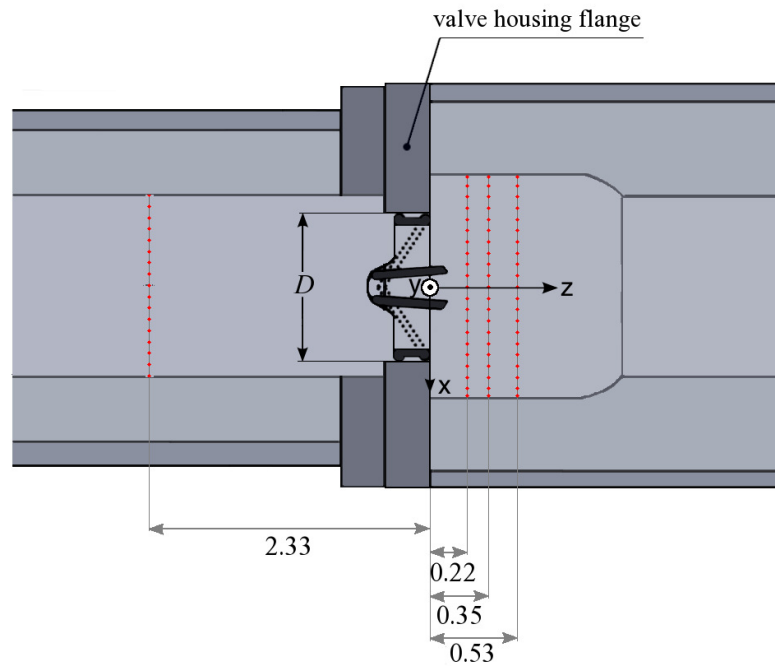
Velocity measurements were recorded for 1000 pulse cycles for each position of the profiles at $z/D = -2.33$ and 4.42 , and for 1500 cycles for each position within the sinus region. A convergence test was made for the measurements of the mean axial velocity $\langle W \rangle$ and the r.m.s. velocity fluctuations w' at location $[x/D = y/D = 0, z/D = 0.35]$. The measurements from the 1500 cycles recorded there were divided into independent subset ensembles of varying cycle number ($M = 150, 250, 375, 500$, and 750). The mean and standard deviations were calculated for each subset, and are plotted in Figure 4.3. As the number of cycles in the ensemble increased, the means $\langle W \rangle$ of the subsets converged to the value obtained from the full ensemble, ($M = 1500$), which we will call $\langle W \rangle_\infty$. The means of the two subsets for $M = 750$ differed, on average, by only 0.2% from $\langle W \rangle_\infty$, indicating that the convergence error for the ensembles with $M = 1500$ was less than 0.2%. The r.m.s. velocity fluctuations also converged with increasing values of M , but not as much as for the mean velocity. The values for the two subsets with $M = 750$ differed on average by 10% of the value at $M = 1500$, called w'_∞ . This difference, however, was only 0.8% of the mean velocity $\langle W \rangle_\infty$ which was much larger than w'_∞ . Based on this, and the infeasibility of increasing the cycle number beyond 1500, it was determined that $M = 1500$ was sufficient for the measurements in the aortic sinus. For the measurements made upstream of the valve at $z/D = -2.33$ and far downstream of the valve at $z/D = 4.42$, the mean velocity was of most interest, as these were made to estimate the flow rate and to establish the symmetry of the boundary conditions. For this reason, 1000 cycles were deemed to be adequate for those locations.

Measurement locations were spaced by 1.6 mm (1/16 in), except at $z/D = -2.33$ and 4.42 , where the space increment was 3.2 mm (1/8 in). Table 4.1 summarizes the coordinates of the measurement locations and the number of cycles recorded for each profile.

The measurements were processed and analysed using the methods described in



(a)



(b)

Figure 4.2: LDV measurement locations in the axisymmetric model (red dots): (a) isometric view; (b) locations near the BMHV on the x - z plane. Dimensions are normalized by $D = 22.6$ mm and referenced from the top of the valve housing flange.

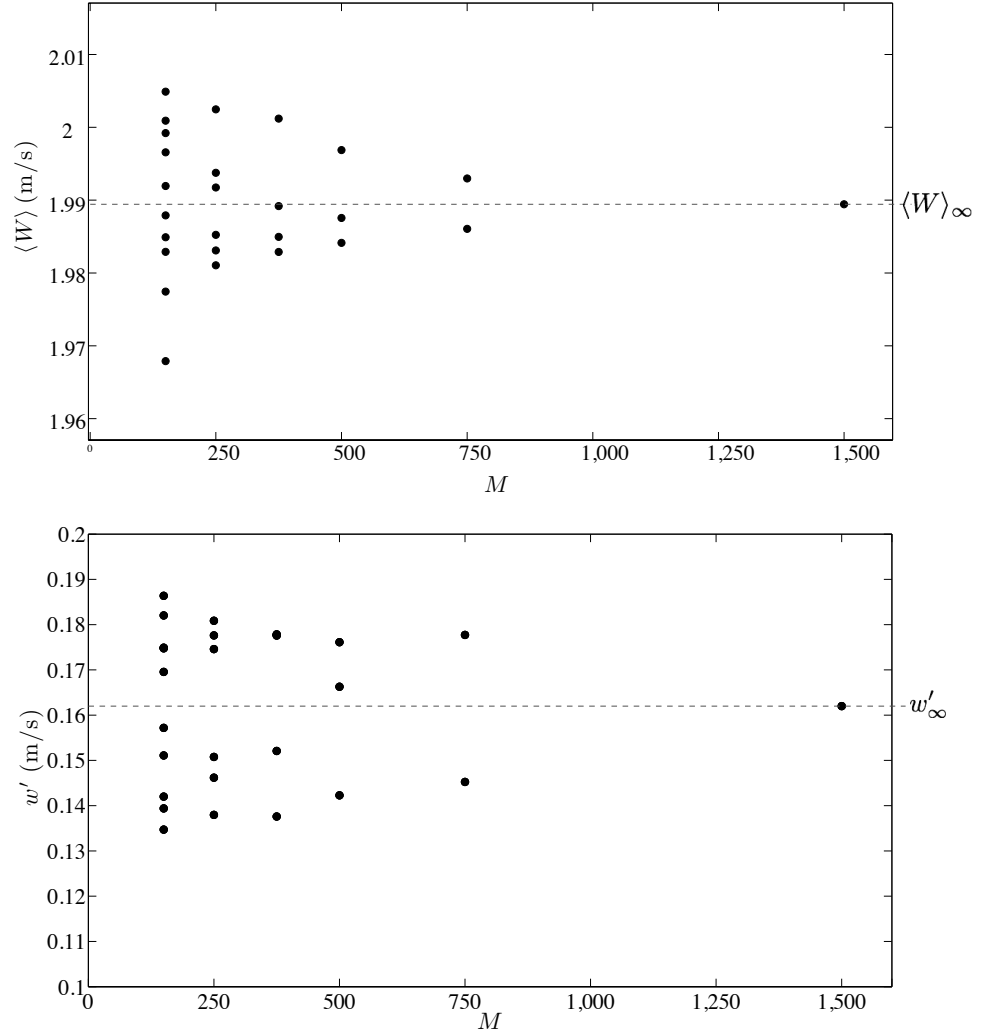


Figure 4.3: Convergence of LDV measurements $\langle W \rangle$ (top) and w' (bottom) for increasing number of acquired cycles M .

Section 3.3.1. The values of the parameters used for processing are listed in Table 4.2. The parameter t_{cutoff} is the maximum number of standard deviations selected for identifying outlying or erroneous data points: the measurement was rejected if the it differed more than this number of standard deviations from the mean. The procedure used for outlier detection is described in Appendix D.

profile	no. cycles	x/D	y/D	z/D
inlet- x	1000	-0.55 : 0.07 : 0.55	0	-2.33
inlet- y	1000	0	-0.55 : 0.14 : 0.55	-2.33
sinus-1- x	1500	-0.69 : 0.07 : 0.69	0	0.22
sinus-1- y	1500	0	-0.69 : 0.07 : 0.69	0.22
sinus-2- x	1500	-0.69 : 0.070 : 0.69	0	0.35
sinus-2- y	1500	0	-0.69 : 0.07 : 0.69	0.35
sinus-3- x	1500	-0.69 : 0.07 : 0.69	0	0.53
sinus-3- y	1500	0	-0.69 : 0.07 : 0.69	0.53
downstream- x	1000	0.55 : 0.07 : 0.55	0	4.42
downstream- y	1000	0	-0.55 : 0.14 : 0.55	4.42

Table 4.1: Coordinates of LDV measurement locations and numbers of cycles recorded.

T (s)	K	Δt (s)	M	$t_{cut-off}$
0.85	180	.0047	1000/1500	5.5

Table 4.2: Values of parameters used for the analysis of LDV measurements.

4.1.2 Inlet flow conditions

The inlet bulk flow rate $\langle Q \rangle_k(z/D = -2.33)$ and aortic pressure P_a are plotted over the normalized cycle time t_k/T in Figure 4.4. The peak flow rate was 24.9 L/min and occurred at $t_k/T = 0.158$. The corresponding pressure at this peak phase was 120.8 mmHg, and represented the systolic pressure. The diastolic pressure, at the beginning of the cycle, was 79.5 mmHg. The aortic pressure curve exhibited an oscillation at the closure of the valve, with a peak pressure of 214 mmHg.

Six key phases of the cycle were identified: ‘A’ labels the start of systole, when forward flow began ($t_k/T = 0.003$); ‘B’ labels the mid-acceleration point ($t_k/T = 0.075$); ‘C’ marks the peak of the forward flow ($t_k/T = 0.158$); ‘D’ corresponds with the mid-deceleration point ($t_k/T = 0.269$); ‘E’ marks peak reverse flow ($t_k/T =$

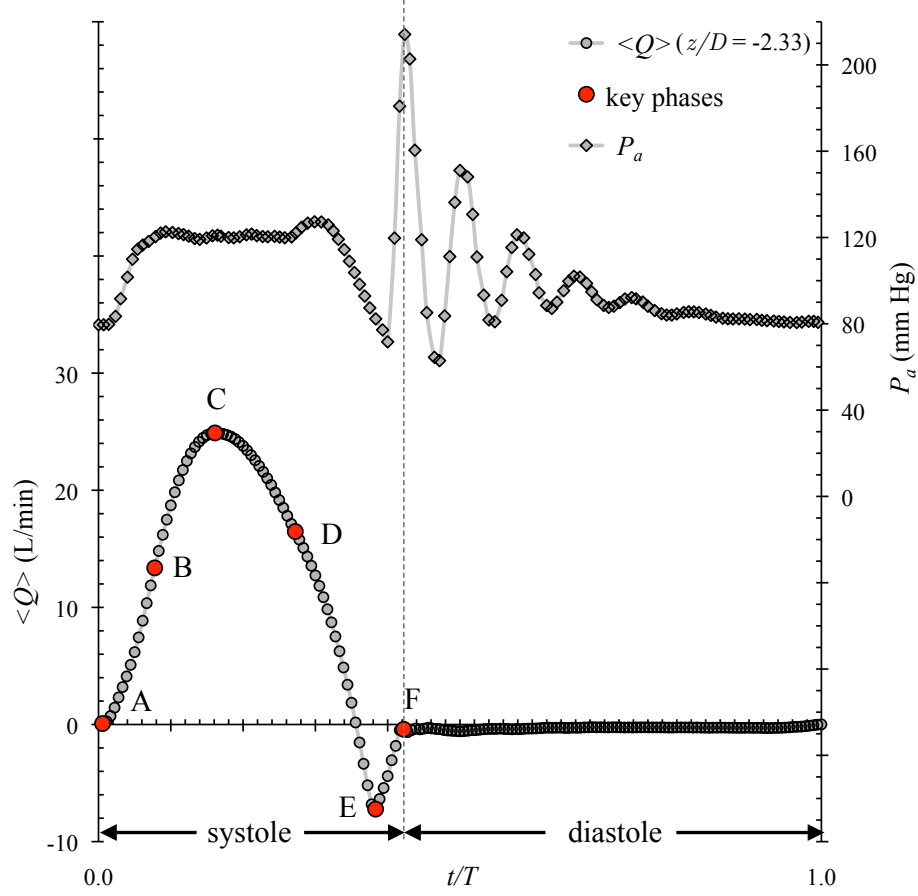


Figure 4.4: Aortic pressure P_a and inlet phase-averaged bulk flow rate $\langle Q \rangle_k$ vs. normalized cycle time t/T ; $T = 0.85$ s. Key phases are labelled: A - start of systole; B - mid-acceleration point; C - peak of the forward flow; D - mid-deceleration point; E - peak reverse flow, and F - valve closure.

0.381); and ‘F’ marks valve closure ($t_k/T = 0.419$).¹ The forward flow persisted for the first 35.3% of the cycle and was followed by a brief period of reverse flow, which lasted 6.6% of the cycle. Diastole persisted for the remaining 58.1% of the total cycle time.

¹It is to be noted that there was a system delay of 85 ms from the time of the solenoid valve opening to the initiation of forward flow and that this and all subsequent flow data was shifted in time by this amount for presentation and referencing purposes. The start of cycle, labeled ‘A’, with t set to equal 0, is defined here by the start of forward flow, and will be referenced as such hereafter.

Nominal z (mm)	$\frac{z}{D}$	$\frac{\overline{Q}_{x-}}{\overline{Q}_{in}}$	$\frac{\overline{Q}_{x+}}{\overline{Q}_{in}}$	$\frac{\overline{Q}_x}{\overline{Q}_{in}}$	$\frac{\overline{Q}_{y-}}{\overline{Q}_{in}}$	$\frac{\overline{Q}_{y+}}{\overline{Q}_{in}}$	$\frac{\overline{Q}_y}{\overline{Q}_{in}}$	$\frac{ \overline{Q}_x - \overline{Q}_y }{\overline{Q}}$	$\frac{\overline{Q}}{\overline{Q}_{in}}$
-53	-2.33	0.99	1.12	1.05	0.77	1.12	0.95	0.11	1.00
5	0.22	—	0.89	0.89	—	0.99	0.99	0.10	0.94
8	0.35	0.81	0.83	0.82	1.15	1.22	1.18	0.37	1.00
12	0.53	0.96	0.61	0.78	—	1.03	1.03	0.27	0.91
100	4.42	0.72	0.76	0.74	0.75	0.84	0.79	0.07	0.77

Table 4.3: Estimates of cardiac output for five axial locations (calculated by Method I, Section 3.3.1): $\overline{Q}_{x-}$, $\overline{Q}_{x+}$, $\overline{Q}_{y-}$, $\overline{Q}_{y+}$, $\overline{Q}_x$, $\overline{Q}_y$, and $\overline{Q}$ normalized by inlet cardiac output $\overline{Q}_{in} = 5.26$ L/min. Axial position z is normalized by the outer diameter of the BMHV, $D = 22.6$ mm. Values for $\overline{Q}_{x-}$ at $z/D = 0.22$ and for $\overline{Q}_{y-}$ at $z/D = 0.22$ and $z/D = 0.53$ were unavailable due to optical limitations preventing complete acquisition of those profiles.

4.1.3 Cardiac output

The cardiac output was calculated for each axial location in accordance with the methods outlined in Section 3.3.1. Table 4.3 summarizes the results: for each location z/D , the normalized solutions $\overline{Q}_{x-}$, $\overline{Q}_{x+}$, $\overline{Q}_{y-}$, and $\overline{Q}_{y+}$ obtained from the four available radii are presented along with the normalized averaged values $\overline{Q}_x$, $\overline{Q}_y$, and $\overline{Q}$. The solutions were normalized by the inlet flow rate $\overline{Q}(z/D = -2.33)$, denoted as $\overline{Q}_{in}$, which was calculated to be 5.26 L/min. The normalized difference between $\overline{Q}_x$ and $\overline{Q}_y$ is presented in the the second last column, and is indicative of the axisymmetry of the flow. As expected, the difference was greatest in the sinus region (37% at $z/D = 0.35$ and 27% at $z/D = 0.53$), where the effects of the valve were greatest. The estimates of $\overline{Q}$ were within the normal physiological range for resting state conditions and ranged between 4.04 and 5.26 L/min.

The hydrodynamic parameters determined for the flow in the axisymmetric test section are summarized in Table 4.4.

fluid density ρ	1230 kg/m ³
kinematic viscosity ν	3.65 cSt
cycle period T	0.85 s
average flow rate	5.26 L/min
peak flow rate	24.9 L/min
forward flow phase	$0.0 \leq t/T \leq 0.35$
Womersley number α	18
average Reynolds number	1200
peak flow Reynolds number	5690

Table 4.4: Hydrodynamic parameters for inlet conditions of the axisymmetric test section. Reynolds and Womersley numbers were calculated using the inlet diameter $d_i = 25.4$ mm.

4.1.4 Velocity profiles

The phase-averaged velocity profiles for $z/D = -2.33$ (inlet) and 4.42 (far downstream) are presented in Figures 4.5a and 4.5b, respectively, at the six key phases of the cycle that were defined in Section 4.1.2. The progression of the profiles can be visualized over the course of the cycle from the left to the right of the figure. The inlet flow demonstrated near plug flow and was nearly axisymmetric, with the exception of small depressions that appeared during the acceleration and peak phases on both negative sides of the x - and y -profiles (Figure 4.5a). The inlet profiles were very organized over the course of the cycle, except at peak reverse flow, when the profile was wavy and disorganized, indicating instability. At valve closure there was minor leakage at the walls.

The downstream flow also demonstrated a plug flow profile and near perfect axisymmetry (Figure 4.5b). The phase-averaged flow profiles were highly organized, even during peak reverse flow. With respect to the upstream coordinate origin, they were slightly convex in shape during acceleration (phase B), and concave in shape during deceleration (phase D). This suggests that the flow initiated from the walls:

the near-wall flow began to accelerate and decelerate prior to the core flow. This was not, however, observed for the upstream profiles ($z/D = -2.33$), for which the profiles were flatter, but somewhat concave at both phases B and D. The upstream profiles were closer to the valve than the downstream profiles, and may have been more influenced by its irregular geometry. Additionally, the wall leakage at valve closure was less for the downstream flow than for the inlet flow, which may again be consequence of the profiles' proximities to the valve. As documented previously by Timité *et al.* [72], the specific shapes of velocity profiles in pulsatile flow are influenced by the Womersley and Reynolds numbers, as well as by the surrounding vessel geometries.

The phase-averaged velocity profiles of the sinus region are plotted in Figures 4.6 and 4.7 for the x - and y -directions, respectively. The x -profiles (Figure 4.6) were perpendicular to the valve's line of symmetry, and illustrated the triple jet generated by the three orifices. The phase-averaged peaks were most prominent near the exit of the valve (at $z/D = 0.22$) and became less prominent further downstream. This was indicative of the dispersion of the jets but may also have been a result of the finite velocity of the pressure wave: upstream fluid was affected by the incoming pressure earlier than fluid further downstream. Portions of the profiles at $z/D = 0.22$ could not be measured due to optical limitations. Values in these portions were estimated as the mirror images of corresponding values measured on the other side and have been shown in Figures 4.6 and 4.7 as grey dashed lines.

The lateral jets formed earlier than the central jet; the prior were visibly strong at mid-acceleration, and the latter had only begun to form at this time. By peak forward flow, the peak velocity of the central jet was 1.99 m/s, and the peak velocity of the lateral jets was about 2.06 m/s. This compares well with the findings of others: Yoganathan *et al.* [86] reported peak velocities of 2.2 and 2.0 m/s for the lateral and central jets at 8 mm downstream of the valve. At the peak flow phase, significant reverse flow occurred in the sinus region near the wall, and persisted through deceleration. This differed significantly from the uniform velocity profiles upstream and far downstream of the valve. In the sinus, narrow, high velocity jets were balanced (in accordance with the conservation of mass) by the reverse flow near the walls. The

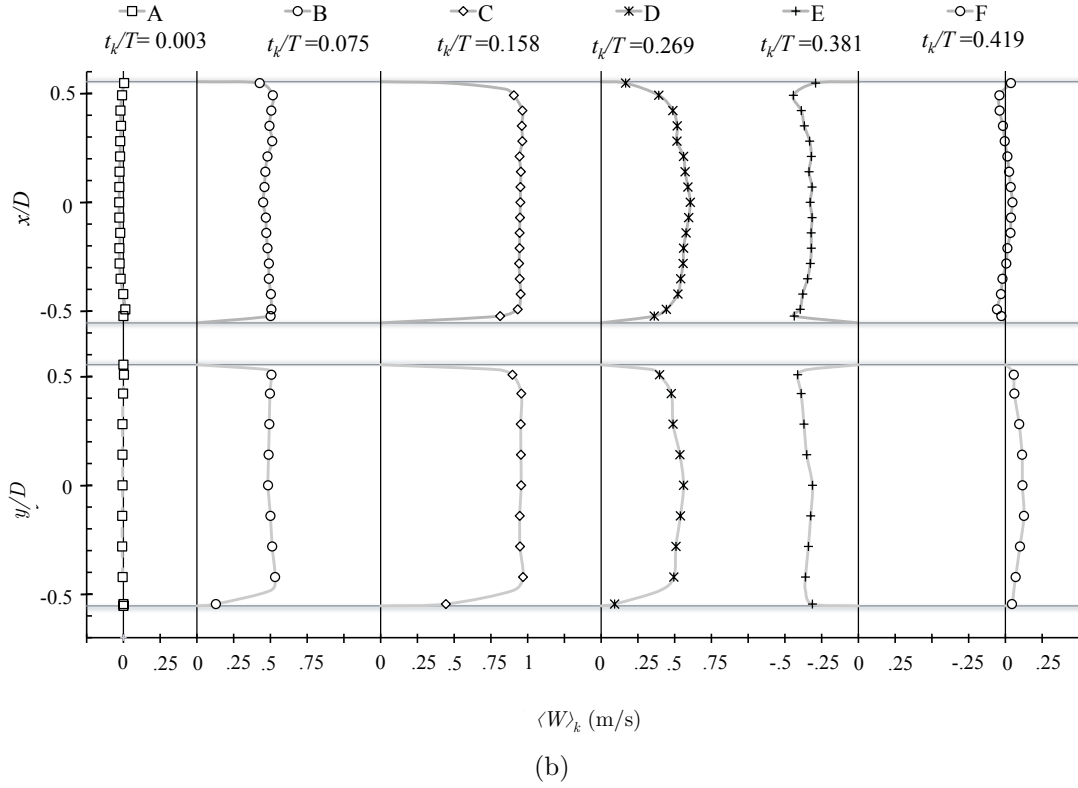
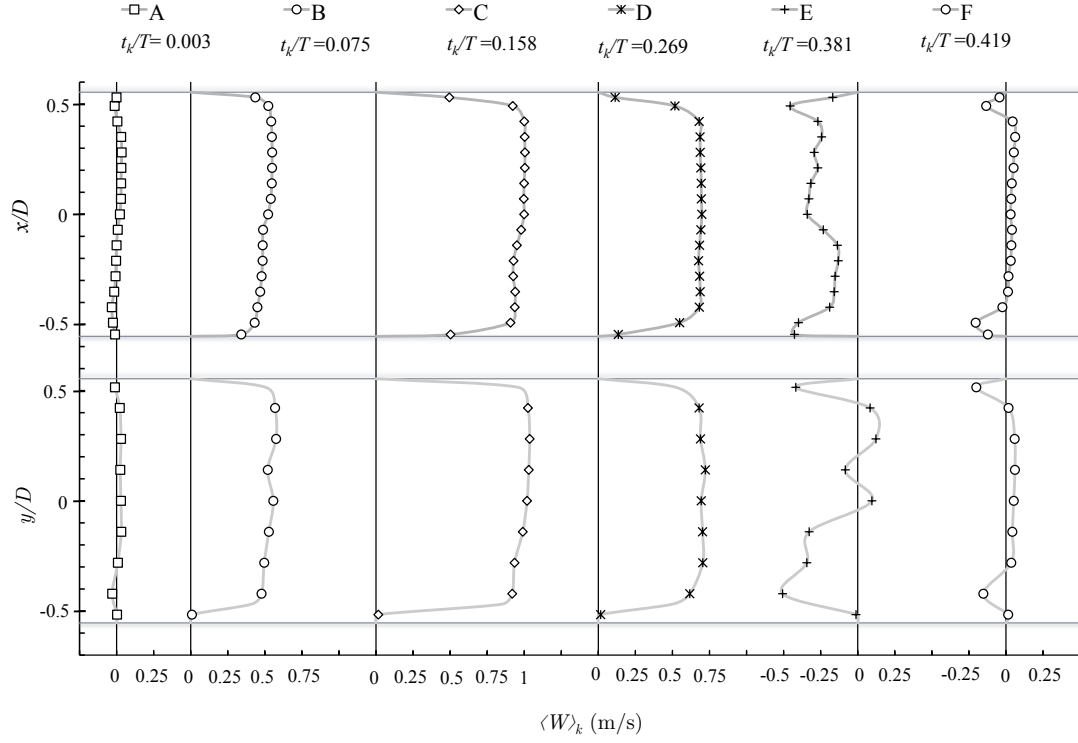


Figure 4.5: (a) Inlet phase-averaged velocity profiles for x (top) and y (bottom) at $z/D = -2.33$. (b) Downstream profiles for x (top) and y (bottom) at $z/D = 4.42$.

peak velocity in the wake of the valve was twice that of the inlet and downstream flows, and very high velocity gradients existed between the lateral jets and the reverse flow regions. At valve closure the velocity profile was bell-shaped in both the x - and y -directions, with small forward flow in the core and small reverse flow near the walls (except at $z/D = 0.53$ in the y -direction, where some disorganization was present in the profile, suggesting that the flow was returning from reverse flow, and at that time had not yet fully restabilized).

The velocity profiles in the y -direction (Figure 4.7) illustrate the side profile of the central jet. The jet initiated from the tube centre, forming a narrow, high velocity jet that transitioned at a “shoulder” (at $y/D \approx \pm 0.28$) to broader, lower speed flow. Some reverse flow occurred near the wall during the peak flow phase and throughout deceleration, but was limited to a much smaller region than was shown by the x -profiles. This indicates that the majority of reverse flow occurred through the lateral orifices and explains why the calculated values of $\overline{Q}$ were consistently greater when calculated from the y -profiles in the sinus than when calculated from the x -profiles.

The shape of the x - and y -profiles closely corresponded with those described by others [10, 20, 52, 75, 86]. In Figure 4.8, the x - and y -profiles at the peak flow phase, at $z/D = 0.35$ are shown with the estimates of de Tullio *et al.* (2011) [75], which were obtained using DNS for two geometries of the aortic sinus: a straight tube and an axisymmetric pseudo-sinus. These profiles were simulated approximately 9 mm downstream of the valve for physiological, pulsatile inlet flow conditions with a cardiac output of 5 L/min. In the x -direction, the velocity profiles of the present results are very similar to those of de Tullio *et al.*, although the peak velocities of the jets were approximately 5 % greater than those of de Tullio *et al.*, and were not as flat. In the y -direction, the jet profile of de Tullio *et al.* was much more narrow than of the present results, while the reverse flow regions were much wider. Despite these differences of distribution, the basic shapes of the y -direction velocity profiles were comparable.

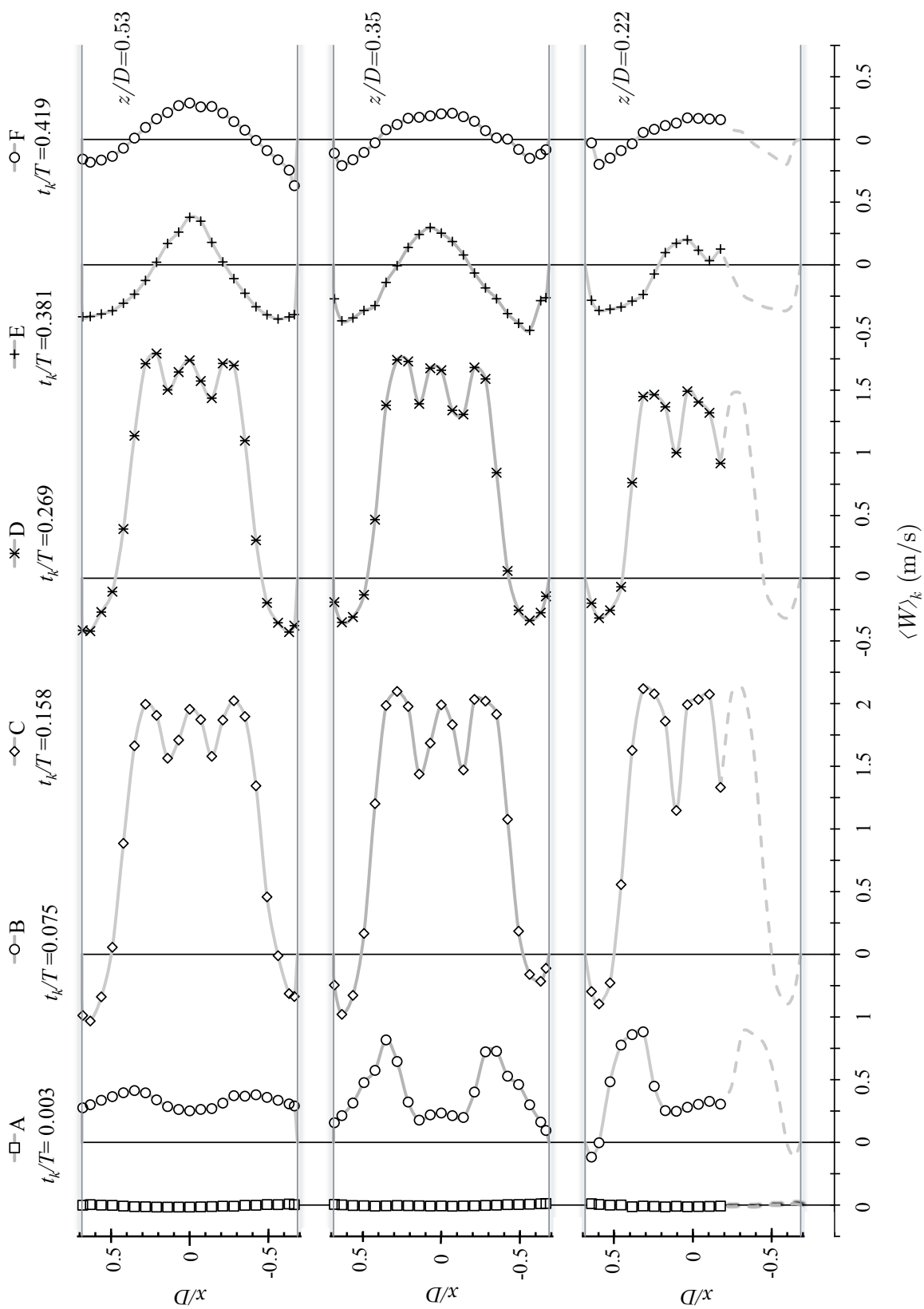


Figure 4.6: Phase-averaged velocity profiles in the x -direction within the axisymmetric sinus, at $z/D = 0.22$ (bottom), 0.35 mm (middle) and 0.53 mm (top). Grey, dashed lines represent portions of the profiles in which measurements could not be taken because of limited optical access to the flow; values in these portions were estimated as mirror images of measurements on the other side of the channel.

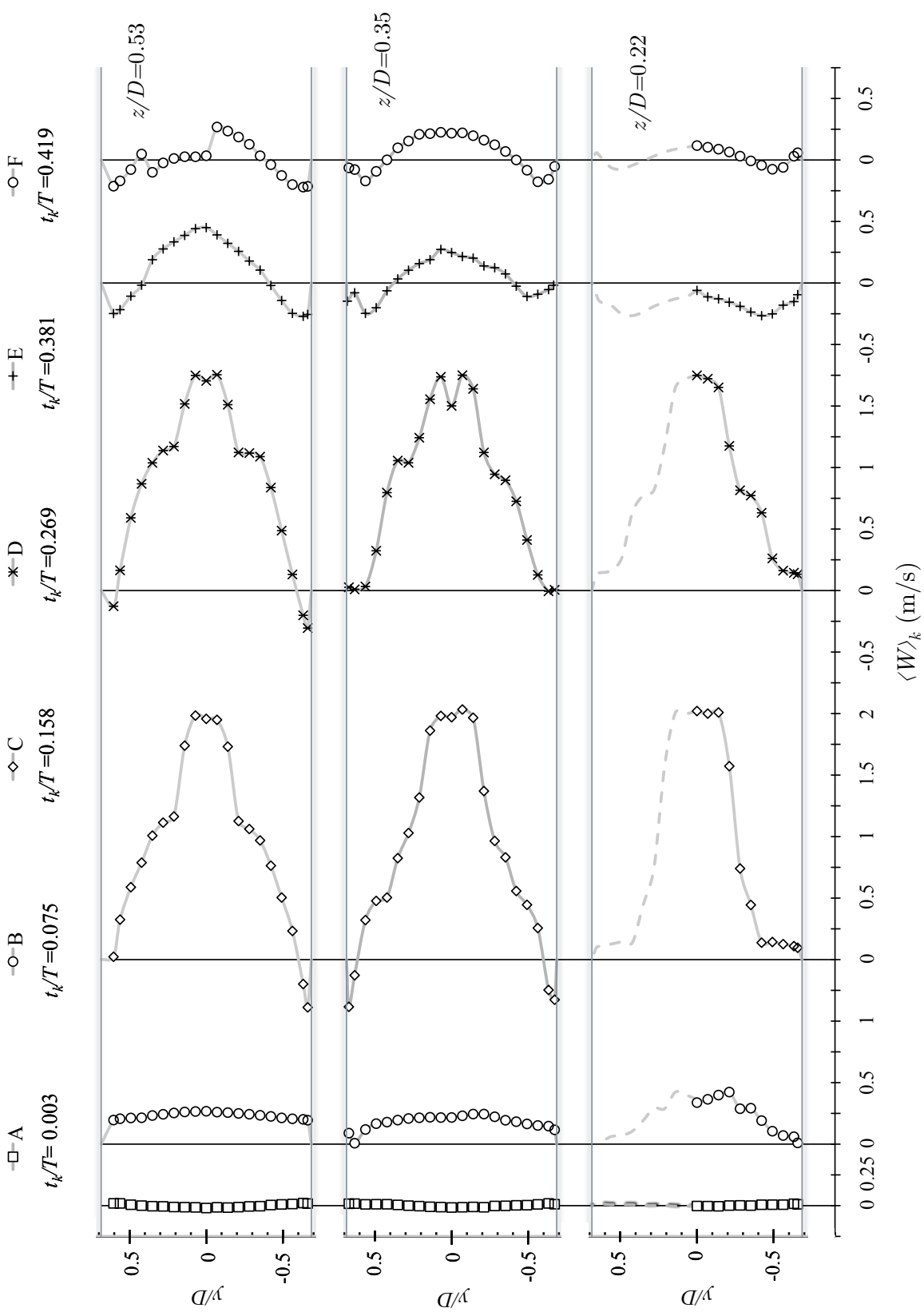


Figure 4.7: Phase-averaged velocity profiles in the y -direction within the axisymmetric sinus, at $z/D = 0.22$ (bottom), 0.35 mm (middle) and 0.53 mm (top). Grey, dashed lines represent portions of the profiles in which measurements could not be taken because of limited optical access to the flow; values in these portions were estimated as mirror images of measurements on the other side of the channel.

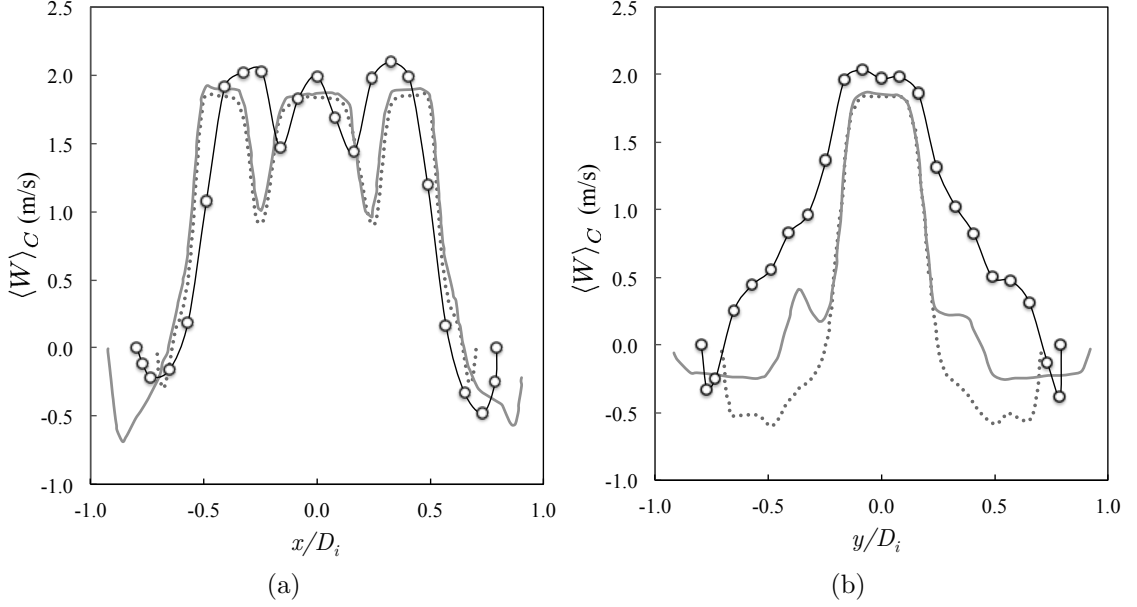


Figure 4.8: Phase-averaged axial velocity at peak flow and $z/D = 0.35$ in the axisymmetric test section: $\circ$ present measurements; DNS results of de Tullio *et al.*[75]: — axisymmetric sinus; straight tube geometry. (a) perpendicular to valve line of symmetry and (b) parallel to valve line of symmetry.

4.1.5 The standard deviations of velocity fluctuations

The turbulent velocity fluctuations w_{ik} were calculated according to Equation 3.6, and the standard deviations of fluctuations for each bin w'_k were then calculated by Equation 3.7. The largest value of w'_k for a given location was denoted w'_{max} . The variation of w'_k over the cycle time was examined for several locations within the sinus, and a time period was identified over which significant fluctuations regularly occurred. This time interval, named the “turbulent period”, was identified between times $t/T = 0.078$ and 0.406 of the cycle, roughly corresponding to the systolic period. In addition to w'_k , which was calculated for each bin, a “systolic-averaged r.m.s.” w'_{sys} was calculated as the standard deviation of fluctuations over this turbulent period. Figure 4.9(a) illustrates the instantaneous velocity W of the 500th cycle for position $[x/D, y/D, z/D] = [0, 0, 0.35]^2$, together with the phase-averaged mean velocity

²This notation will hereafter be used to indicate coordinates.

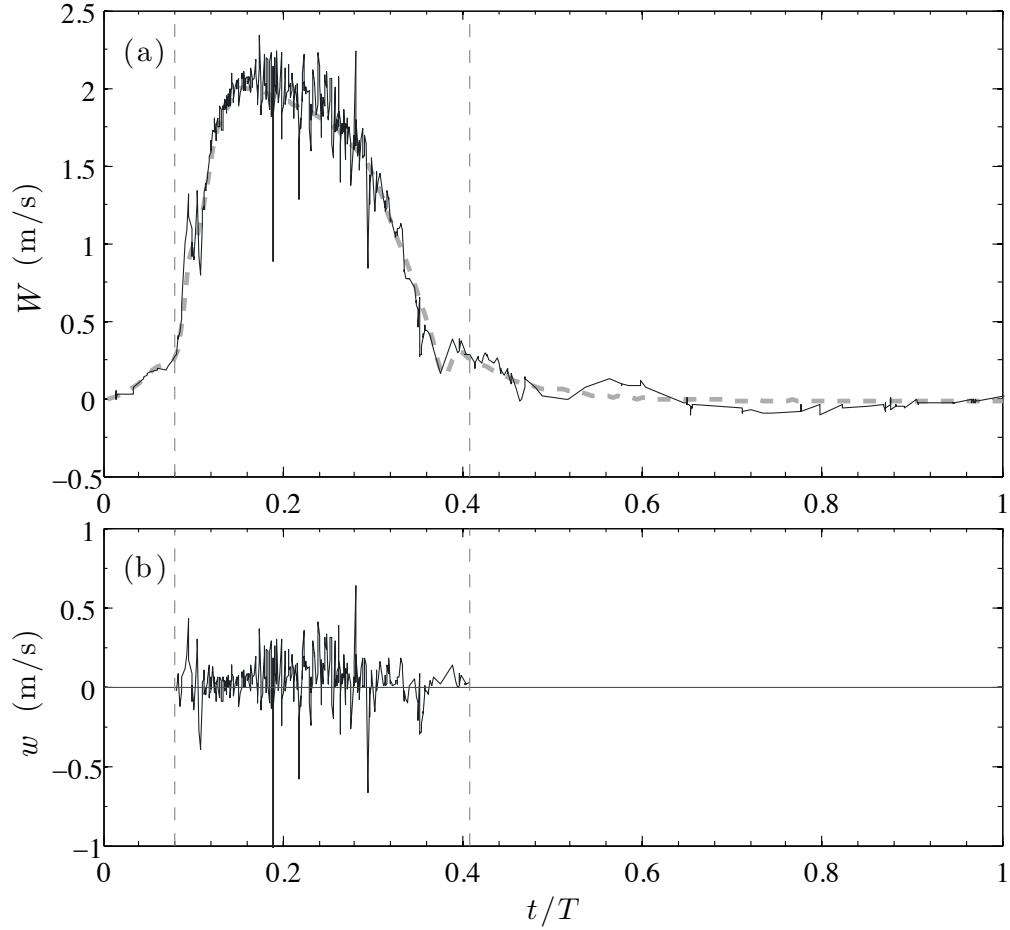


Figure 4.9: From LDV measurements at location $[0,0,8]$ in the axisymmetric test section: (a) Instantaneous velocity W of the 500th cycle (solid line), with the phase-averaged velocity $\langle W \rangle_k$ (heavy, dashed line); (b) Velocity fluctuations w of the 500th cycle. The turbulent period is enclosed by the dashed vertical lines.

$\langle W \rangle_k$. The fluctuations w_{ik} , obtained from the removal of the mean from the instantaneous measurements, are shown for the turbulent period in Figure 4.9(b). The fluctuations became large just prior to peak flow and fluctuated roughly about zero over the course of the turbulent period.

Figures 4.10 and 4.11 illustrate the variation of w'_{sys} and w'_{max} in the x - and y -directions, respectively, within the sinus. The profiles are presented for the three sinus axial locations ($z/D = 0.22$ 0.35, and 0.53), and the corresponding velocity profiles of the peak flow phase $\langle W \rangle_C$ are shown for reference.

For all locations, the w'_{max} profile closely followed the same shape as the corresponding w'_{sys} profile, but was, on average, 50% greater. By superimposing the w' profiles onto the peak velocity profiles, it is evident that the r.m.s. fluctuations were greatest at regions of high velocity gradients. In the x -direction, this corresponded with the outer sides of the lateral jets, and to a lesser extent, the sides of the central jet. The peaks of w' at the sides of the central jet were most prominent nearest the valve ($z/D = 0.22$), and smoothed to insignificance further downstream ($z/D = 0.53$). In the y -direction, the highest levels of w' corresponded with the steepest sides of the central jet, again, correlating with the velocity gradient. Despite the high velocity of the central jet, a depression of w' existed at the location of the jet's flat peak, which was likely a result of the near zero velocity gradient at this location.

The greatest value of w'_{max} calculated for all measured locations past the valve was 0.532 m/s, occurring at location $[-0.42, 0, 0.35]$, and $t/T = 0.178$ (shortly after peak flow). The profile average of w'_{sys} slightly increased with downstream progression, from 0.206 m/s at $z/D = 0.22$, to 0.233 m/s at $z/D = 0.53$ in the x -profiles, and similarly from 0.208 to 0.236 m/s in the y -profiles. In summary, velocity fluctuations past the valve were on average about 0.22 m/s over a turbulent period of about 300 ms, while maximum values were typically slightly higher than 0.5 m/s and normally occurred at the onset of deceleration.

These calculated values of velocity fluctuations were comparable, but slightly greater than those determined by Liu *et al.* [52], who, by phase-averaging, found w' to be 0.265 m/s at location $[0.09, 0, 0.34]^3$, and 0.353 m/s at location $[0.43, 0, 0.34]^3$, past the valve at the peak phase of the flow⁴. Comparative values from this study were determined to be 0.287 m/s (at $[0.06, 0, 0.35]$) and 0.488 m/s (at $[0.42, 0, 0.35]$). Although the two studies operated under similar cardiac outputs of approximately 5 L/min, the peak velocity in the present study was 25% greater than that in the study by Liu *et al.* and the systolic duration was 13% shorter; these differences resulted in greater temporal velocity derivatives for the present measurements, which

³The equivalent position using the coordinate convention of the present study. The coordinates from Liu *et al.* [52] were normalized by $D = 23$ mm.

⁴The peak flow phase of the study by Liu *et al.* [52] corresponded with the three phase bins centred about the peak flow phase of the present study.

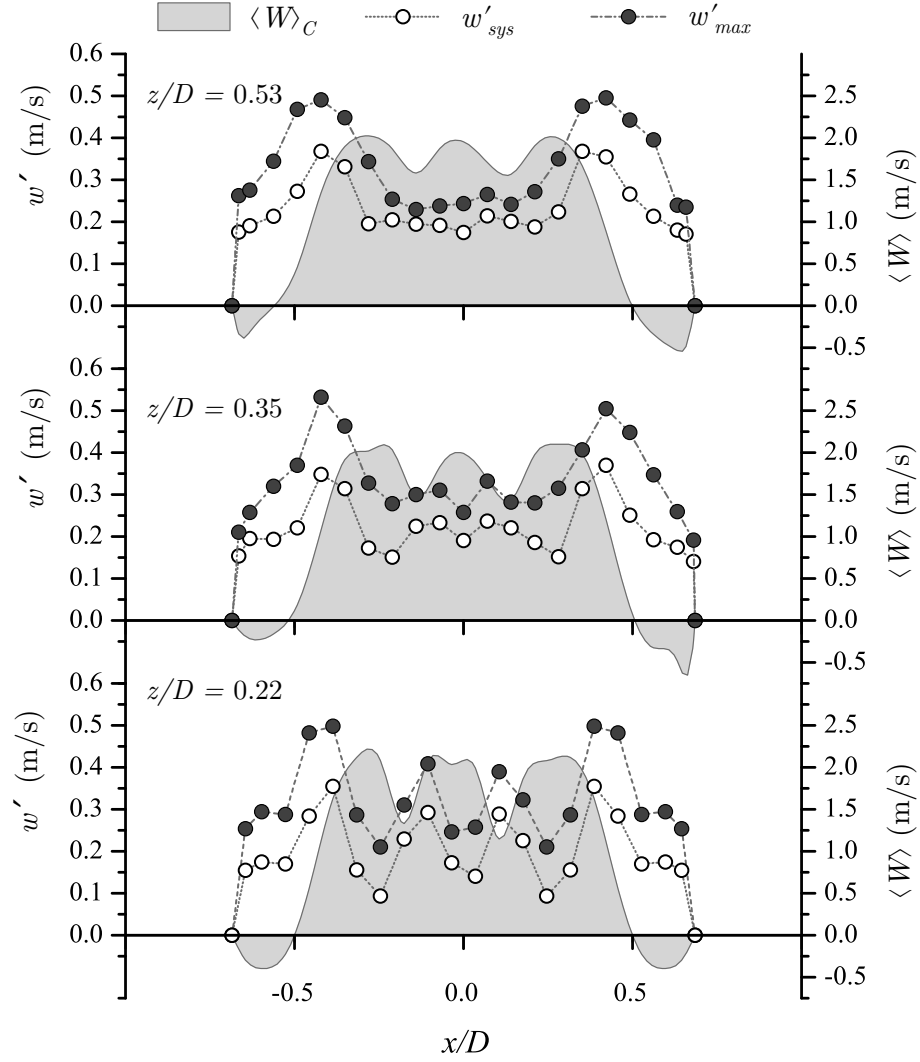


Figure 4.10: Standard deviations of velocity fluctuations w'_{max} (solid markers) and w'_{sys} (open markers) for the x -direction, at three axial locations: $z/D = 0.22$ (bottom), 0.35 (middle), and 0.53 (top). Phase-averaged axial velocity $\langle W \rangle_C$ (peak flow) is shaded in grey for reference.

may have accounted for the slightly higher values of velocity fluctuations.

4.2 Planar velocity measurements

4.2.1 Measurement locations and sampling conditions

For the planar velocity measurements in the axisymmetric model, the bileaflet valve was mounted in the same position as for the single-point measurements: with its plane

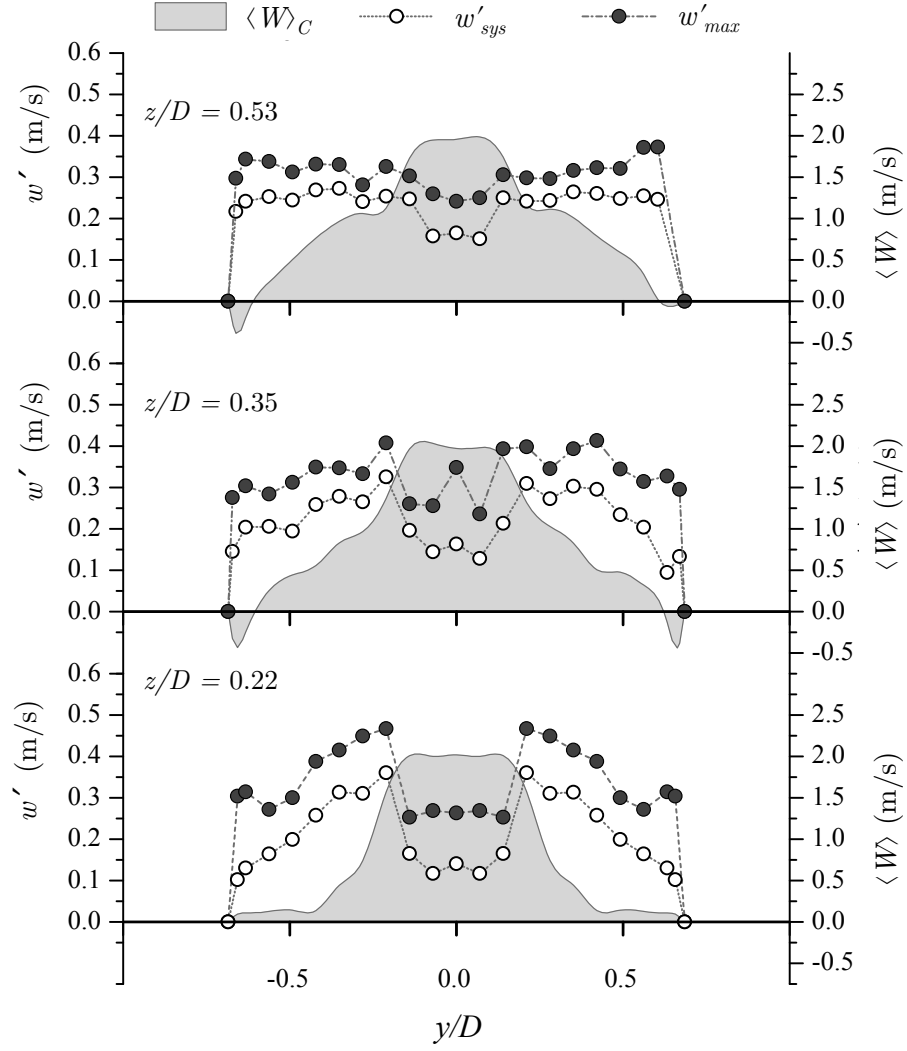


Figure 4.11: Standard deviations of velocity fluctuations w'_{max} (solid markers) and w'_{sys} (open markers) for the y -direction, at three axial locations: $z/D = 0.22$ (bottom), 0.35 (middle), and 0.53 (top). Phase-averaged axial velocity $\langle W \rangle_C$ (peak flow) is shaded in grey for reference.

of symmetry corresponding with the coordinate system's y -axis. The mock circulation loop was adjusted to deliver pulsatile flow to the test section having the same pressure and flow rate conditions as were used for the LDV measurements (shown in Figure 4.4).

Phase-locked, 2D PIV measurements were made downstream of the valve in planes parallel and perpendicular to the valve's plane of symmetry (Figure 4.13). In the x - z and y - z centre-planes, 250 instantaneous flow maps were acquired at each of 27 time

instances over the forward-flow phase of the cycle (shown as the “other phases” in Figure 4.14). These phases corresponded to every third time bin to which the LDV measurements were sorted. The number of measurements acquired was increased to 1000 for selected key phases (Table 4.5).

A convergence test was made for the measurements of the mean axial velocity $\langle W \rangle$ and the r.m.s. velocity fluctuations w' at location $[x/D = y/D = 0, z/D = 0.5]$, in the same way as was done for the LDV measurements (Figure 4.12). For the ensembles with $M = 500$, the values of $\langle W \rangle$ and w' differed from $\langle W \rangle_\infty$ and w'_∞ on average by 0.1% and 3%, respectively; this indicates that for the ensembles with $M = 1000$, the convergence error was less than 0.1% for the mean velocity and less than 3% for the r.m.s. velocity fluctuations. For the ensembles with $M = 250$, the convergence error was less than 0.2% for the mean velocity, and less than 6% for the r.m.s. velocity fluctuations.

In addition to the centre-plane measurements, measurements were made in planes corresponding to the locations of maximum velocity fluctuations, as determined from the LDV measurements made past the BMHV (at $x/D = -0.421$ and at $y/D = -0.211$). At each of these planes, denoted as “max-r.m.s.” planes, 1000 measurements were acquired at the time of maximum fluctuation, $t/T = 0.181$. The vectors for these and the centre-plane measurements were calculated using a 12×12 pixel interrogation window with a 50% window overlap. The resulting vector resolutions were $105 \mu\text{m}$.

The details of all PIV measurements including locations, phases, numbers of cycles recorded, and vector resolutions are summarized in Table 4.5.

4.2.2 Planar velocity maps

Instantaneous and phase-averaged velocities in the x - z centre-plane of the axisymmetric model are presented in Figures 4.15 and 4.16 for phases A,B,C and phases D,E,F, respectively. The images in the left-hand column are sample instantaneous measurements, while the images in the right-hand column are the phase-averages of 1000 measurements.

The instantaneous flow images at the very start of the cycle (phase A), were

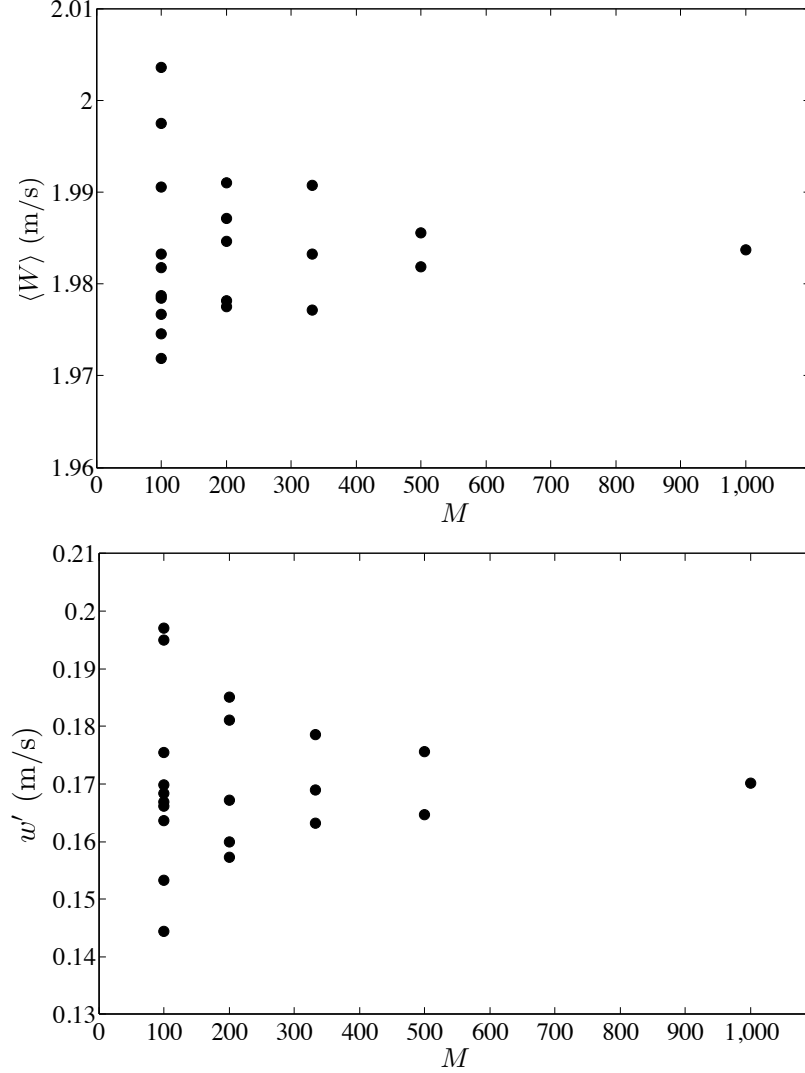


Figure 4.12: Convergence of PIV measurements $\langle W \rangle$ (top) and w' (bottom) for increasing number of image pairs M .

characterized by low velocity and random vortices, which were not apparent in the averaged field. The average flow was very slow (maximum velocity less than 0.02 m/s); it was downward in the channel centre and upward near the walls. In agreement with previous observations [86], the lateral jets were more developed and stronger than the central jet at mid-acceleration (phase B). They were flat-topped, with clearly defined boundaries. Recirculation zones formed between the lateral jets and the channel walls, and flow separation was apparent at the leaflet tips. The instantaneous image

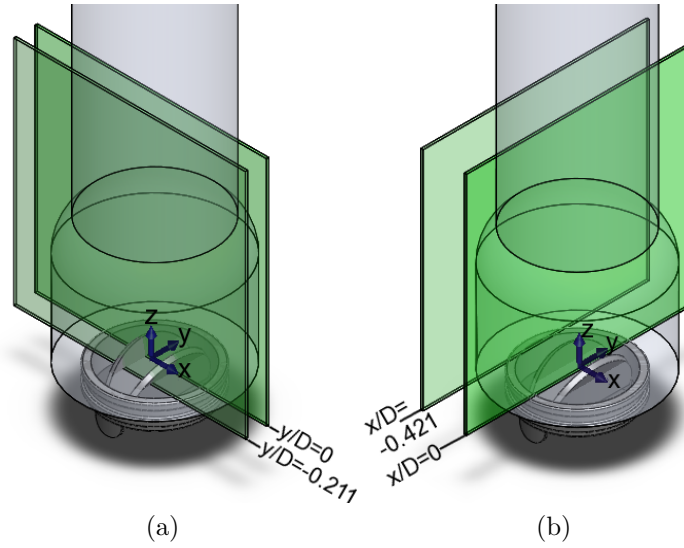


Figure 4.13: PIV measurement planes in the axisymmetric test section: (a) x - z planes; (b) y - z planes

plane	out of plane value	region	phases	no. cycles	resolution (μm)
x - z	$y/D = 0$	sinus	$t/T=0.003 : 0.0167 : 0.436$	250	104
x - z	$y/D = 0$	sinus	A, B, C, D, E, F	1000	104
x - z	$y/D = -0.211$	sinus	$t/T = 0.181$	1000	105
y - z	$x/D = 0$	sinus	$t/T=0.003 : 0.0167 : 0.436$	250	104
y - z	$x/D = 0$	sinus	B, C, D, E	1000	104
y - z	$x/D = -0.421$	sinus	$t/T = 0.181$	1000	105

Table 4.5: Planes, phases, numbers of cycles recorded, and vector resolutions for PIV measurements made in the axisymmetric test section.

matched very closely the averaged image, indicating that the flow at this phase was highly organized and repeatable. At peak flow (phase C), the jets extended past the end of the sinus. The waviness of the instantaneous jets indicated instability and turbulence associated with high velocities. The sinus recirculation in the instantaneous field was less coherent than at phase B, but in the averaged field, the recirculation zones appeared as coherent, elongated vortices, with their centres of circulation located at the end of the sinus. The random waviness of the instantaneous jets was

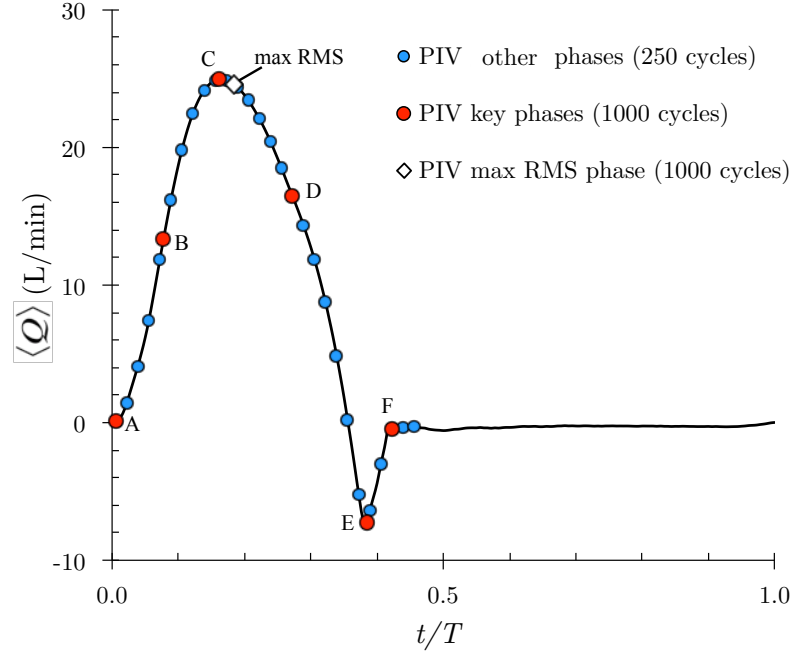


Figure 4.14: PIV phases indicated on the bulk flow rate waveform.

not apparent in the averaged field, which depicted the jets as being smooth, with the lateral ones stronger and extending further downstream than the centre one. All flow features, including the jets, were very symmetric about the z -axis ($x = 0$).

Figure 4.16 shows velocity maps for the mid-deceleration phase (D), peak reverse flow (E) and valve closure (F). At mid-deceleration the flow appeared to be unstable and random. The lateral jets were more inward than at phase C, and the sinus recirculation zones were larger and dominated the flow field. The instantaneous features of the triple jets mixed, and the jets appeared to have merged at approximately 0.75 diameters downstream of the valve. At peak reverse flow (phase E) the jets were fully retreated, and reverse flow occurred near the walls of the channel with a maximum velocity of about 0.8 m/s. Very low velocity upward flow occurred in the channel core. The instantaneous velocity at valve closure (phase F) showed random vortices with peak velocities of 0.45 m/s, which were not visible when averaged. Like at phase E, the averaged flow was upward in the core and downward at the walls, whereas the centre flow was not very different from that at phase E (about 0.25 m/s), the downward wall flow was substantially lower (less than 0.15 m/s).

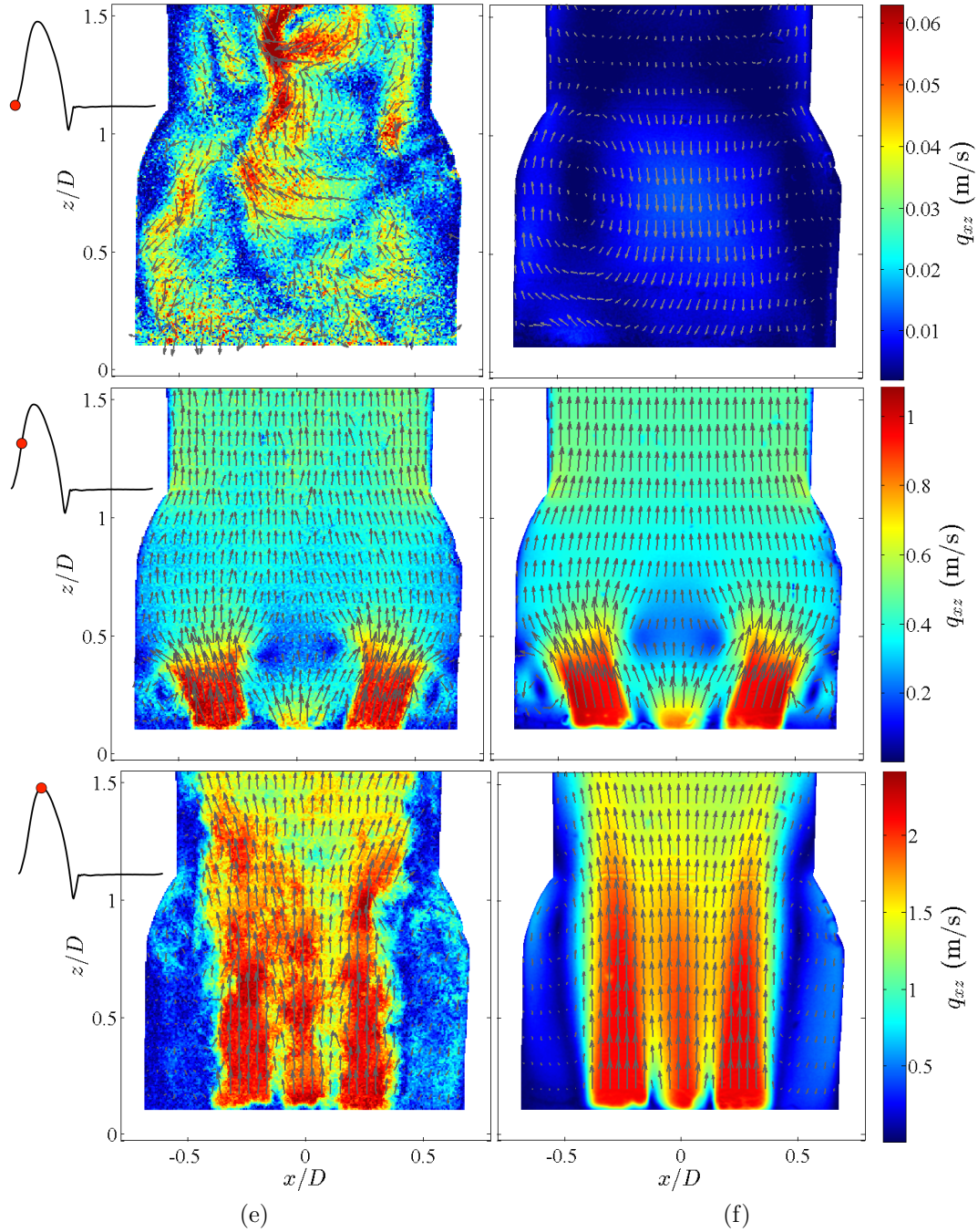


Figure 4.15: Planar velocity q_{xz} in the axisymmetric model (perpendicular to valve plane of symmetry, $y = 0$). Left column: instantaneous velocity; right column: phase-averaged velocity (sample size 1000). Top row: phase A ($t/T = 0.003$); middle row: phase B ($t/T = 0.075$); bottom row: phase C ($t/T = 0.158$).

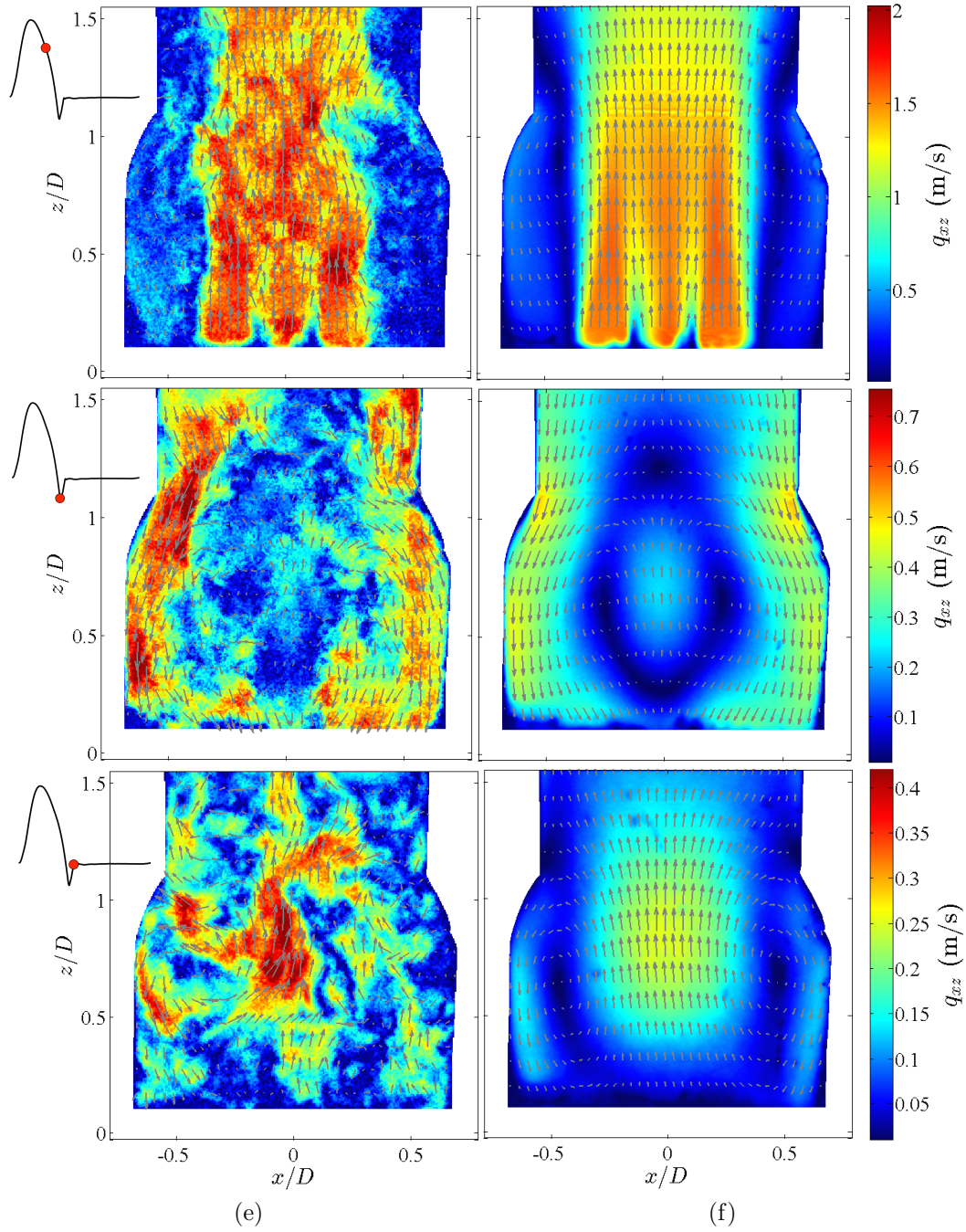


Figure 4.16: Planar velocity q_{xz} in the axisymmetric model (perpendicular to valve plane of symmetry, $y = 0$). Left column: instantaneous velocity; right column: phase-averaged velocity (sample size 1000). Top row: phase D ($t/T = 0.269$); middle row: phase E ($t/T = 0.381$); bottom row: phase F ($t/T = 0.419$).

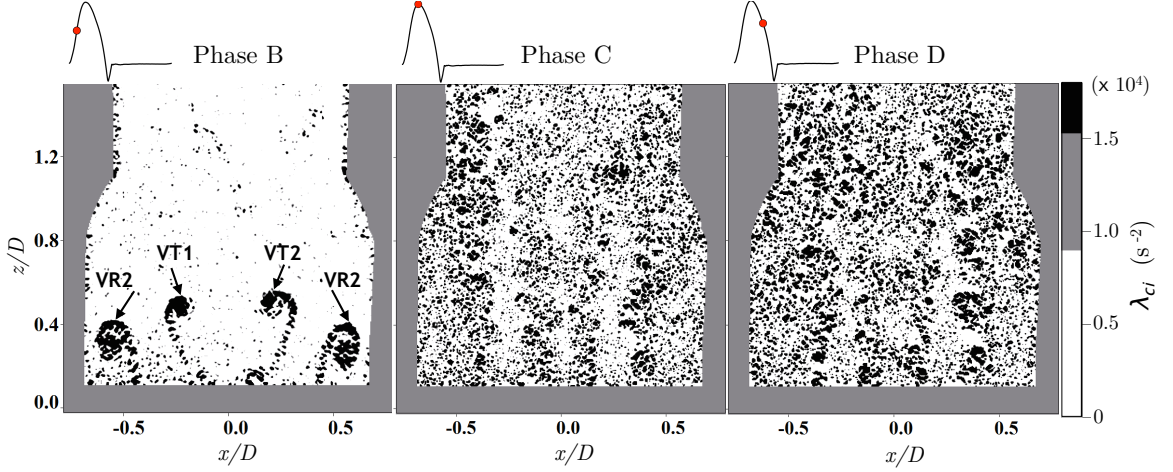


Figure 4.17: Swirling strength λ_{ci} on the x - z centre-plane of the axisymmetric test section for phases B (left), C (middle) and D (right).

Figure 4.17 shows the swirling strength λ_{ci} for sample instances at mid-acceleration, peak flow, and mid deceleration. At mid-acceleration (phase B), four distinct vortices were visible: one shed from each leaflet tip (respectively labelled VT1 and VT2) and one shed from each side of the valve housing (labelled VR2) which occupied the sinus between the valve housing and the wall; these latter ones are in fact the 2D projections of a single vortex ring, which Dasi *et al.* [27] called the “sinus ring”. The vortices in Figure 4.17 are labelled by the same notation used by Dasi *et al.* [27], and can be compared with the ones identified by their 3D simulation, shown previously in Figure 2.7. By peak flow (phase C), these structures were no longer visible, and the swirling strength map was characterized by many small, random vortices, indicating that the flow was turbulent. The swirling strength map for the mid-deceleration phase D was very similar to that for peak flow: it was also characterized by many small random vortices, indicating that turbulence persisted. For both the peak and mid-deceleration phases, the turbulent vortices were slightly less concentrated at the three streaks corresponding to the velocity jets; they were more concentrated at the sides of the jets, where the velocity gradient was steep.

Phase-averaged and instantaneous velocity profiles are plotted for the y - z plane in Figure 4.18 for phases B, C, D and E. They characterize the flow of the central jet. At phase B, the jet barely emerged past the valve leaflets. At phase C, the jet

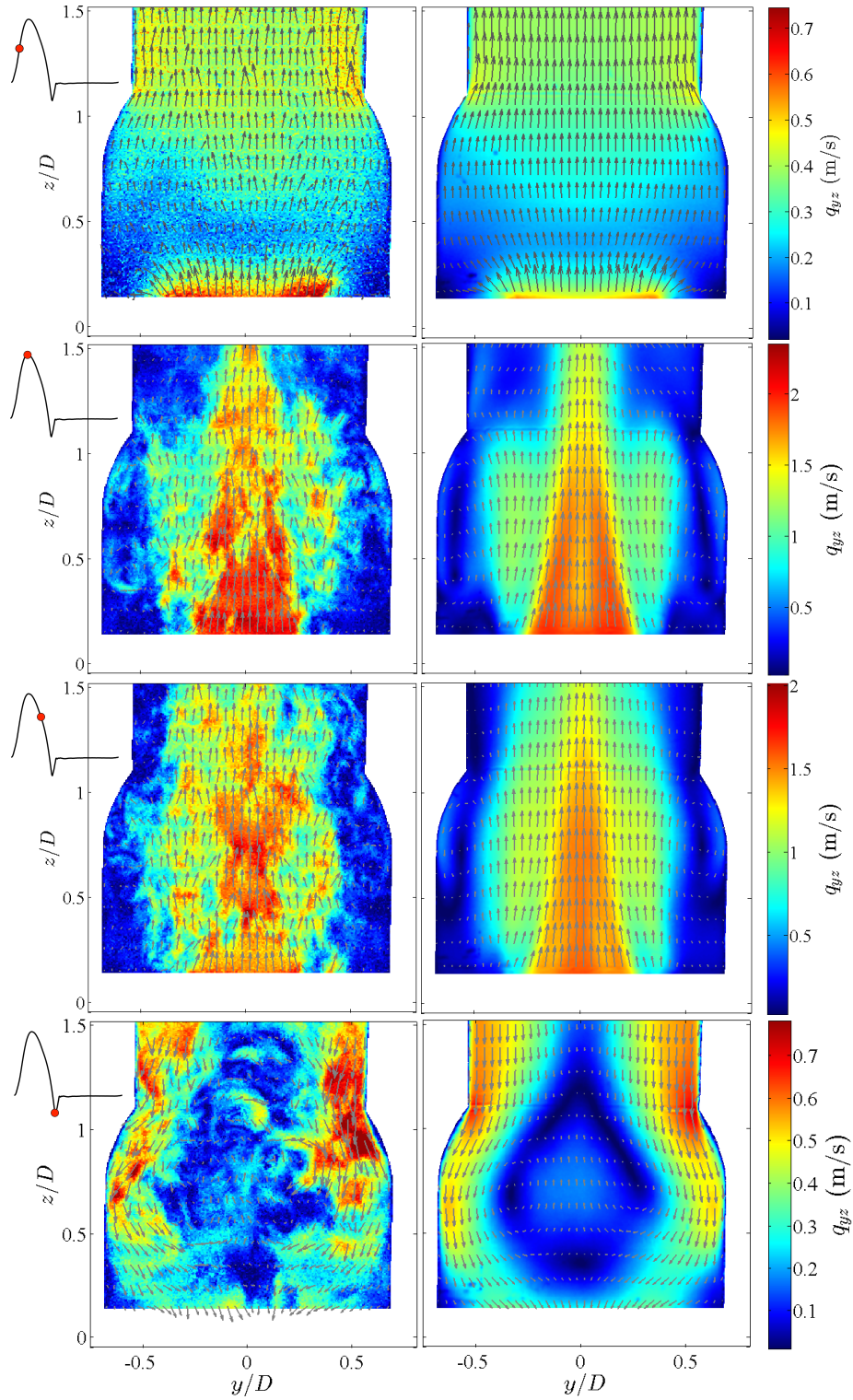


Figure 4.18: Planar velocity q_{yz} in the axisymmetric model (parallel to valve plane of symmetry, $x = 0$). Left column: instantaneous velocity; right column: phase-averaged velocity (sample size 1000). Top row: phase B ($t/T = 0.075$); second row: phase C ($t/T = 0.158$); third row: phase D ($t/T = 0.269$); bottom row: phase E ($t/T = 381$).

was strong and turbulent and extended past the end of the pseudo-sinus; it narrowed downstream, and a mid-range velocity ‘shoulder’ bridged the high velocity jet centre and the sinus recirculation flow. The averaged field revealed a lower velocity ‘dimple’ at the jet centre, which was similar to the one measured by Browne *et al.* [14]. Oblong recirculation regions formed in the sinus, with reverse flow at the walls. For clarity, we shall refer to certain features of the centre-jet referred as the ‘shoulder’, the ‘dimple’, and the ‘head’; these are illustrated in Figure 4.20 (c). At phase D, the averaged jet strength diminished, but the instantaneous field revealed the flow was more random than at peak flow, with many high velocity packets formed throughout the sinus. The velocity at phase E closely resembled that of the same phase in the x - z plane: the retrograde flow washed down the channel walls, with a low upward velocity region occupying the core of the channel.

4.2.3 Comparison with previous measurements and with LDV results

The phase-averaged axial velocity $\langle W \rangle$ and r.m.s. velocity fluctuations w' were compared with profiles by Yun *et al.* [88], obtained numerically with the entropic lattice-Boltzmann method (LBM) and by Dasi *et al.* [27], obtained with PIV. The profiles are plotted in Figure 4.19 for peak flow, for position $z/D_i = 0.234$ downstream of the valve ($D_i = 21.4$ mm for both previous studies). The velocities were normalized by W_0 , the average velocity of the corresponding profile (1.012 m/s for the present study and 0.720 m/s, for both previous studies). The shapes of the jet profiles were nearly identical, although the normalized peak jet velocities of the present study were on average 8% greater, and the central jet was more rounded than that of Yun *et al.* [88] (Figure 4.19(a)). The reverse flow in the x^- sinus closely matched the results of Dasi *et al.* [27], but, in the x^+ sinus, better matched those of Yun *et al.* [88]. The normalized velocity fluctuations (Figure 4.19 (b)) were also in very good agreement with those reported by Yun *et al.* [88] in both shape and magnitude. The largest discrepancy was of the fluctuations corresponding with the outer shear layer of the x^- jet, in which those of the present study were larger by 31%. Overall, the mean and

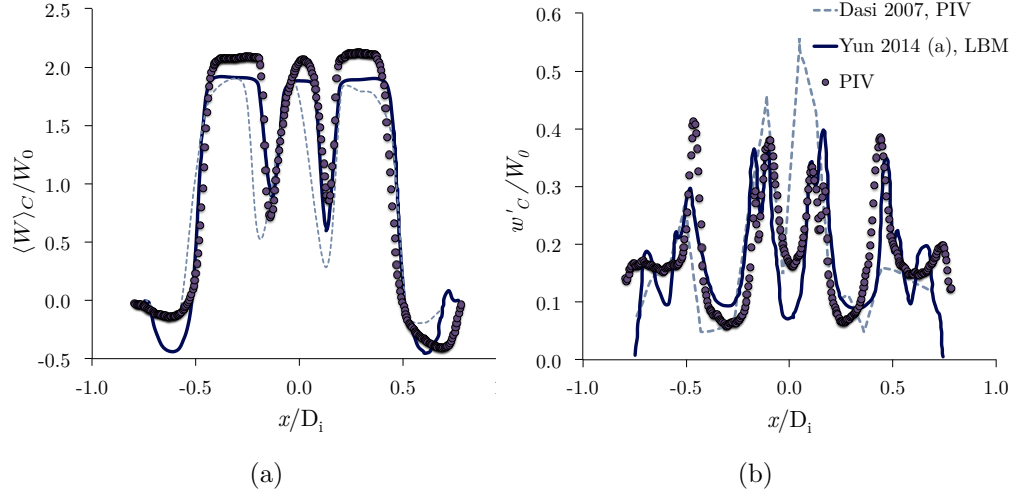


Figure 4.19: Phase-averaged (left) and standard deviation of fluctuations (right) of axial velocity at peak flow, and $z/D_i = 0.234$, in the axisymmetric channel, compared with similar measurements by Yun *et al.* [88] and by Dasi *et al.* [27].

r.m.s. velocities showed excellent agreement with other measurements under comparable conditions, which inspires confidence in the results presented herein.

In Figure 4.20, the results from the PIV measurements are plotted with those from LDV for the x -axis and the y -axis at downstream location $z/D = 0.53$, at peak flow. The averaged axial velocity is shown on the left and the r.m.s. axial velocity fluctuations on the right. Overall, there was good agreement between the two methods. In some locations, the velocity fluctuations were predicted higher by LDV than for PIV, most notably at the outside edges of the lateral jets and in the sinuses (Figure 4.20 (b)). This may have been due to a particle sample bias that can occur in shear layers as a result of lower seeding densities within them and, consequentially, lower sampling rates [15]. The velocity fluctuations measured from LDV were also higher at the sides of the jet head in the y -plane, but were lower at the middle of the jet (Figure 4.20(d)). This is likely because, unlike for the PIV measurement, the central jet measured by LDV did not exhibit a ‘dimple’ in its middle (Figure 4.20(c)). This is partly attributed to the separation by several months of the LDV and PIV measurements, which may have introduced a change in flow conditions. This may also have resulted from the accumulative differences between the two methods;

for example, there may have been a slight mis-alignment of the measurement plane or location for either method. The presence of this ‘dimple’ in the central jet, as shown by the PIV measurements, would be associated with slightly higher velocity fluctuations due to the imposed velocity gradients, as compared with the flat-topped jet. Overall, the measurements made with LDV closely matched those made with PIV, further supporting our confidence in the results.

It should be noted that the application of the Gaussian smoothing filter to the vector fields in post-processing caused a reduction in the r.m.s. velocity fluctuations: for the x - z centre-plane at peak flow w' was decreased on average by 8%. Figure 4.21 shows the results of w' calculated from the vectors before and after Gaussian smoothing, for the profile at $z/D = 0.5$, plotted with the results from the LDV measurements.

4.2.4 Turbulent stresses

Maps of the maximum turbulent stress RS_{max} are presented in Figure 4.22. Regions of greatest RS_{max} corresponded to the outer edges of the lateral jets, specifically, the x^+ -side jet, where values up to 170 Pa were reached. This maximum value of RS_{max} is very close to the value 180 Pa reported by Ge *et al.* [35]. The regions of high turbulent stress extended past the end of the sinus, up to the jet reattachment points. On the x - z centre-plane, the RS_{max} increased in intensity downstream of the valve, peaking at about $z/D = 1.0$. On the max-r.m.s. plane, however, the stress was greatest immediately downstream of the valve and dissipated further downstream, although it sustained values as high as 100 Pa. The RS_{max} was substantially less on the y - z centre-plane with maximum values around 100 Pa but confined to relatively small areas. Values were greater on the y - z max-r.m.s. plane with peak values of 140 Pa located at the sides of the narrow central jet. Wide regions of RS_{max} up to 100 Pa persisted downstream, past the end of the sinus.

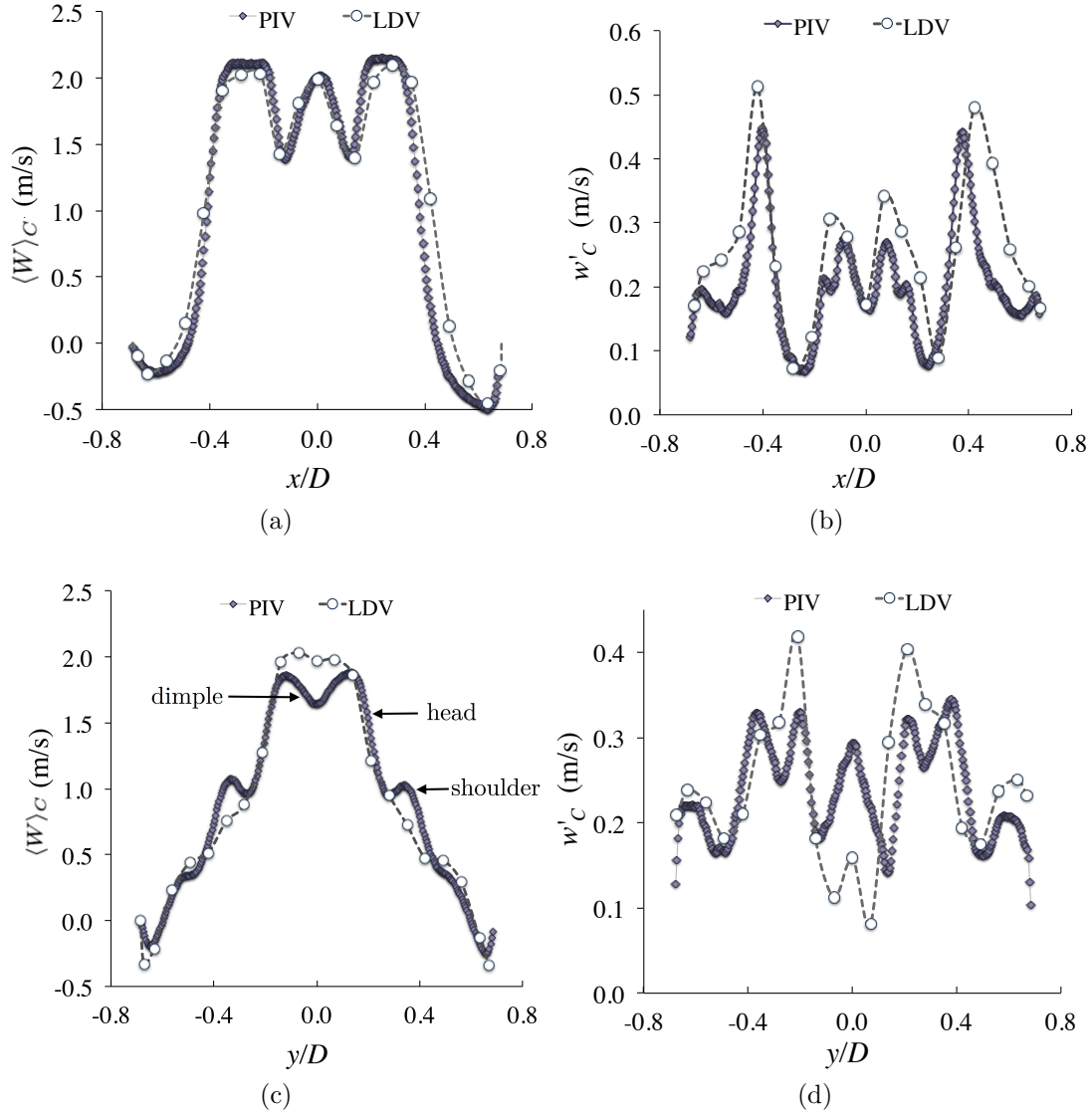


Figure 4.20: Comparison of measurements made with PIV to those made with LDV: phase-averaged (left) and r.m.s. velocity fluctuations (right) of axial velocity at peak flow, and at $z/D = 0.53$, in the axisymmetric channel, along the x -axis (top) and the y -axis (bottom).

4.2.5 Viscous shear stresses

Maps of viscous shear stress τ_{max} are presented in Figure 4.23 for the axisymmetric model on the x - z and y - z centre-planes and the max-r.m.s. planes. The regions of high shear stress corresponded closely with those of high turbulence, most notably in the shear layers at the sides of the jets. The stress was highest immediately downstream

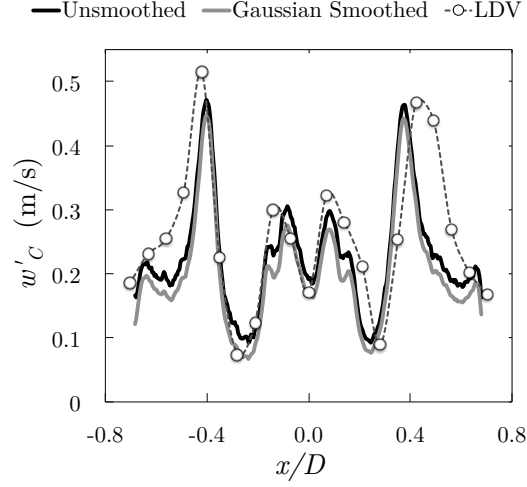


Figure 4.21: Effect of Gaussian smoothing on w' measured with PIV.

of the valve, and the regions of high stress widened downstream. Instantaneous values of the viscous shear stress were as high as 20 to 30 Pa on the x - z centre-plane (Figure 4.23(a)), but such instances were infrequent, and the maximum stress in the averaged field was 15 Pa (Figure 4.23(b)). This corresponded with the location directly off of the leaflet tips, where flow separation occurred and vortices were shed. The levels of shear stress were higher on the x - z max-r.m.s. plane, where the regions of high shear stress were widespread. The maximum stress on the averaged field was 16 Pa.

The viscous shear stress in the y - z planes was, on average, much less than on the x - z planes. The averaged maximum stress was 6 Pa on the centre-plane and 9 Pa in the max-r.m.s plane. The regions of maximum stress corresponded to the sides of the central jet. Again, compared with the centre-plane, the overall levels of viscous shear stress on the max-r.m.s. plane were higher and the regions of high stress were more widespread.

To show the viscous shear stress distribution for each plane, histograms were generated from the measurements of 1000 scalar fields. They were normalized by the area beneath them ($A = \Delta\tau N$, where $\Delta\tau$ is the bin width and N is the total number of measurements) to give probability density functions (PDFs), denoted $f(\tau_{max})$ (Figure 4.24). The probability that any measurement would have been within a given range of τ_{max} is equal to the integral of $f(\tau_{max})$ over that range.

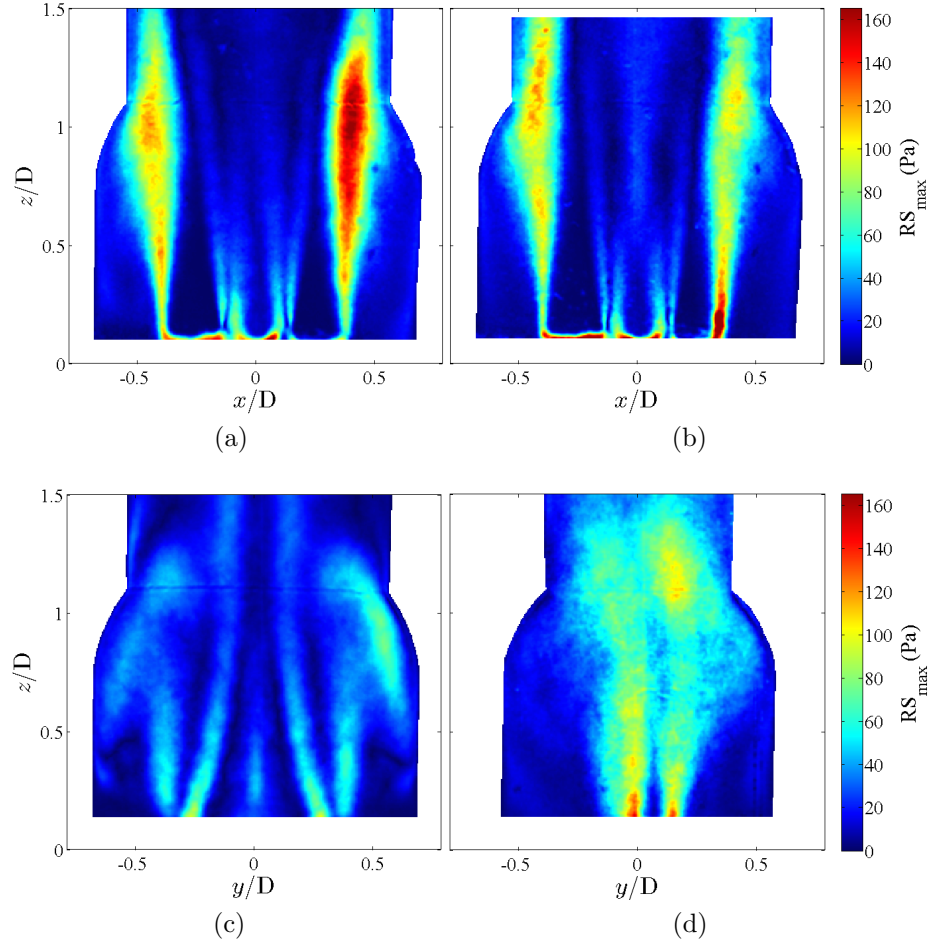


Figure 4.22: RS_{max} calculated in the axisymmetric model at peak flow for: (a) the x - z centre-plane; (b) the x - z max-r.m.s. plane; (c) the y - z centre-plane; and (d) the y - z max-r.m.s. plane.

It has been previously shown that viscous shear stress values as low as 10 Pa can cause damage to platelets [35]. The PDFs show that the majority of measured stress in all cases was less than 10 Pa and that a very small percentage was greater than 18 Pa. The x - z max-r.m.s. plane contained the highest levels of stress, with 11.6% of measured values exceeding 10 Pa. It was followed by the y - z max-r.m.s. plane, having 6.9% of measurements exceeding 10 Pa, and the x - z centre-plane and the y - z centre plane with 4.0% and 0.3% of measured values exceeding 10 Pa, respectively.

For their 3D simulations past a BMHV in an axisymmetric channel, Borazjani *et al.* [12] reported that, at a phase just prior to peak flow, 0.02% of measurements

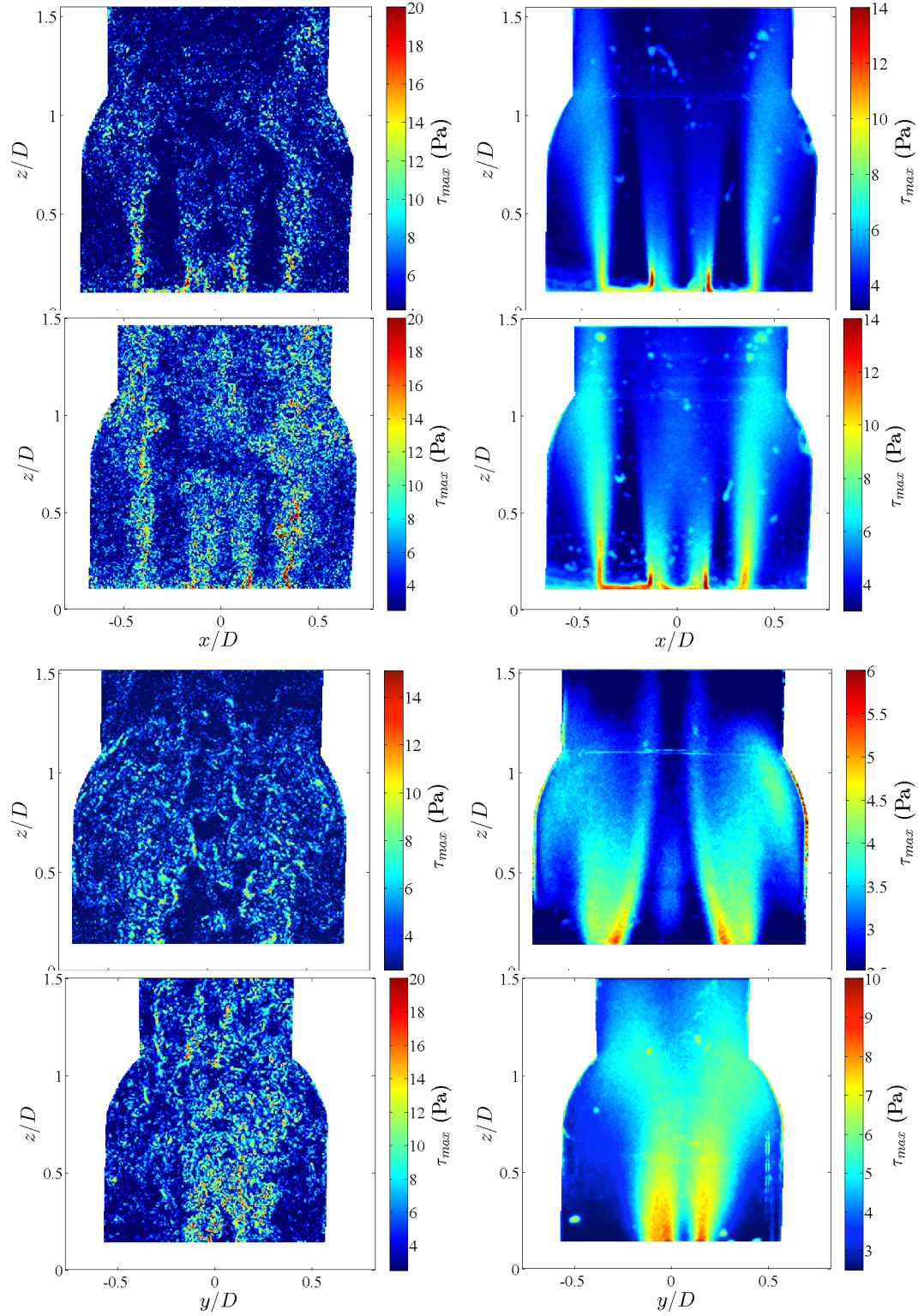


Figure 4.23: Instantaneous (left) and phase-averaged (right) values of τ_{max} in the axisymmetric model in the x - z centre-plane (top), the x - z max-r.m.s. plane (2nd row), the y - z centre-plane (3rd row) and the y - z max-r.m.s. plane (bottom).

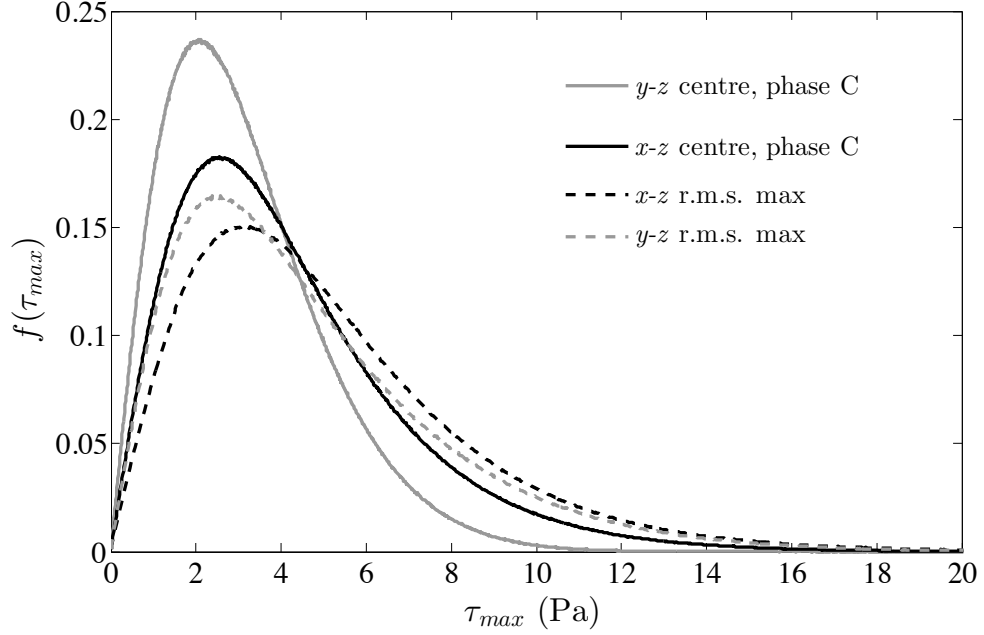


Figure 4.24: Probability density functions (PDFs) for viscous shear stress in axisymmetric model for x - z and y - z centre-planes at phase C, and x - z and y - z planes and phases for “max-r.m.s.”

corresponded to the bin at $\tau_{max} = 10$ Pa in a histogram with $\Delta\tau = 0.064$ Pa; the normalized PDF value was $f = 0.003$. In the present study, we also found that $f = 0.003$ for $\tau_{max} = 10$ Pa on the y - z centre-plane; however, on the x - z centre-plane we found that $f = 0.018$ for $\tau_{max} = 10$ Pa — a value six times larger. It is noted that the histograms reported by Borazjani *et al.* [12] applied to the entire three-dimensional domain past the valve, whereas in the present study only individual 2D planes were considered. It is expected that high levels of viscous shear stress would be concentrated in certain regions, such as near the jet edges, and that large regions of the flow domain would have relatively low levels of viscous shear stress. It is therefore unsurprising that the PDF at high values of τ_{max} was many times greater for measurements on planes in high-stress regions than for the full 3D domain.

4.3 Analysis of the results and discussion: Axisymmetric model

The turbulent stress metric RS_{max} was considerably higher on both x - z planes (centre and “max-r.m.s.”) than either of the y - z planes (Figure 4.22). This was consistent

with the fact that the x - z planes transected all three jets and the shear layers between them, where large spatial velocity gradients and velocity fluctuations occurred; whereas the y - z planes were parallel to the leaflets and did not cross the multiple shear layers that formed in their wake. Additionally, there was more space in the y -direction for the jets to spread, which reduced the region of strong velocity gradient.

As also expected, RS_{max} was higher on the y - z max-r.m.s. plane than on the y - z centre-plane (Figure 4.22). The y - z max-r.m.s. plane transected the outer shear layer of the x^- -side lateral jet, occupying a region of very strong spatial velocity gradient. The y - z centre-plane, however, bisected the central jet, where the phase-averaged velocity variation in the x -direction was negligible and, as a result, the flow had lower fluctuations.

On the other hand, and rather unexpectedly, RS_{max} was higher in extensive regions on the x - z centre-plane than on the x - z max-r.m.s. plane (Figure 4.22). This can be attributed to two factors:

1. The location of the x - z max-r.m.s. plane was determined from LDV measurements taken at just two profiles across the centre of the channel (at $z/D = 0.22$ and $z/D = 0.35$), neglecting measurements in large regions of the flow. The chosen plane, therefore, did not necessarily contain the highest velocity fluctuations of the entire domain.
2. Velocity profiles in the y - z max-r.m.s. plane, which transected the x^- -side lateral jet, reveal that the jet was skewed toward the y^+ wall (Figure 4.25). The distribution of velocity fluctuations was similarly skewed: w' was slightly greater at $y/D = 0$ the location of the x - z centre-plane than at $y/D = 0.211$ (the location of the x - z max-r.m.s. plane) (0.47 m/s compared to 0.43 m/s, respectively). As a result, the x - z centre-plane did not coincide with the peak of this lateral jet but transected its y^- edge, where velocity fluctuations were higher due the relatively strong velocity gradient.

The turbulent stress is indirectly correlated with the velocity gradient but is also affected by the velocity magnitude and other factors affecting turbulence, including the temporal rate of change of velocity. This is reflected in the results, in which the

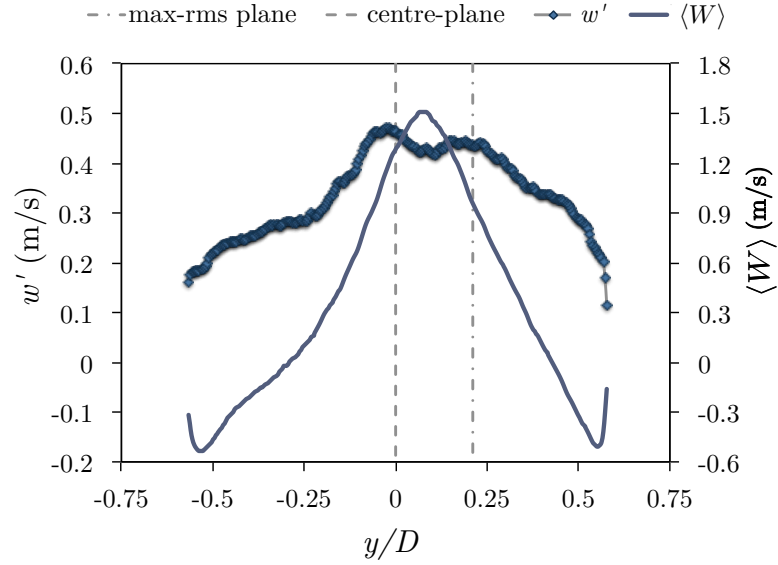


Figure 4.25: Phase-averaged axial velocity and velocity fluctuations at $z/D = 0.75$ on the y - z max-r.m.s. plane in the axisymmetric model.

highest levels of turbulent stress seem to be correlated with but do not exclusively correspond to the locations of greatest velocity gradient.

Conversely, the viscous shear stress is by definition proportional to the spatial velocity gradient and independent of the velocity magnitude and rate of change. In consequence, the highest levels of viscous shear stress seem to correspond to the planes and regions of highest velocity gradient. For example, in all measurement planes high levels of viscous shear stress τ_{max} were located at the edges of the jets.

Chapter 5

Results: Velocity Measurements in an Anatomical Aorta Model

5.1 Measurement locations and sampling conditions

The aorta model was inserted in the mock circulation loop with its plane of curvature approximately parallel with the x - z plane. The non-coronary sinus was located toward the negative side of the y -axis, and the right and left coronary sinuses were located on the negative and positive sides of the x -axis, respectively (note the convention of the “anatomical right” appearing on the left-hand side of diagrams, and vice versa). The x - z and y - z centre-planes are shown on the aorta model in Figure 5.2. The masks used for PIV processing are indicated on this figure by red lines, outside of which vectors were not calculated. The valve was mounted at the inlet of the aorta model, first, in the same position relative to the coordinate system as it was in the axisymmetric channel; *i.e.*, with its plane of symmetry aligned with the y -axis (Figure 5.1). In this position, the right and left coronary sinuses were situated approximately symmetrically about the valve’s plane of symmetry. The outlets of all branching arteries were fully clamped so there was no flow through them. In this chapter, the results of this “basic configuration” will be presented and compared to the measurements made in the axisymmetric channel. The results for other valve orientations with and without flow through the right coronary artery will be presented in Chapter 6.

The loop was adjusted to generate pressure and flow conditions matching as closely as possible those used for the axisymmetric measurements; however, due to the necessary reconfiguration of the loop to accommodate the aorta model, it became impossible to reproduce exactly the shapes of the pressure and flow waveforms. The aortic pressure P_a measured downstream of the valve and the bulk flow rate $\langle Q \rangle$ are plotted

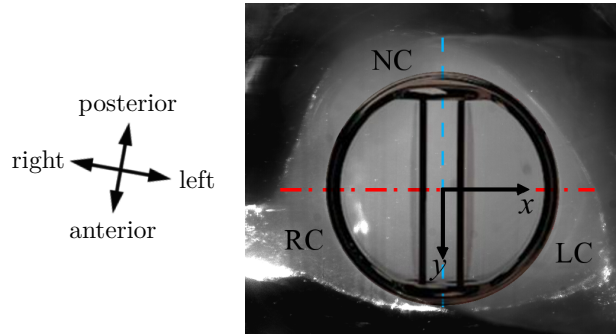


Figure 5.1: Top-down image of BMHV positioned within the anatomical sinus: dash-dot red line: x - z centre-plane; dashed blue line: y - z centre-plane; RC: right coronary sinus; LC: left coronary sinus; NC: non-coronary sinus. Approximate anatomical directions are indicated by the compass.

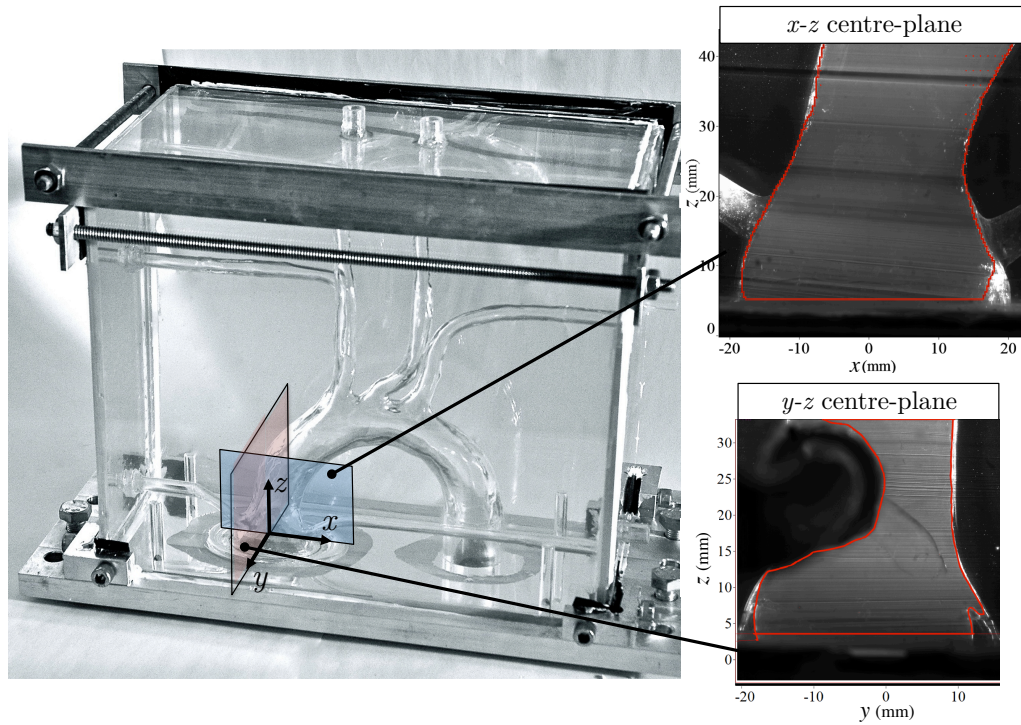


Figure 5.2: Anatomical model of aorta showing x - z and y - z PIV planes with averaged particle images. Red lines mark the masks, outside of which vectors were not calculated.

	A	B	C	D	E	F	max-r.m.s.
Axisymmetric	0.003	0.075	0.158	0.269	0.381	0.419	0.181
Anatomical	0.003	0.108	0.214	0.314	0.414	0.428	0.225

Table 5.1: Key phase times (t/T) for measurements in the axisymmetric and anatomical models.

in Figure 5.3 for flows in both the axisymmetric and anatomical models. As for the flow in the axisymmetric model, key phases were defined at the start of the cycle (A), mid-acceleration (B), peak flow (C), mid-deceleration (D), peak reverse flow (E) and valve closure (F); however, the times of the key phases were slightly different for the two models. The results will be compared for matching key phases rather than cycle times. The times of the key phases, normalized by the cycle period T , are tabulated in Table 5.1 for both the axisymmetric and anatomical measurements.

Again, measurements were made on the x - z and y - z centre-planes, as well as on off-centre planes corresponding to the locations of maximum velocity fluctuations (at $x/D = -0.421$ and at $y/D = -0.211$). At each of these “max-r.m.s.” planes, 1000 measurements were acquired at the times of maximum fluctuation ($t/T = 0.225$ for the anatomical model). The details of the PIV measurements made in the anatomical model for the “basic configuration” including locations, phases, and numbers of cycles recorded are summarized in Table 5.2.

5.2 Planar velocity maps

Several explanations must be given about the anatomical measurements before presenting the results:

1. A circular area in the y - z field, located at the top of the non-coronary sinus, was masked from viewing, because this area was optically blocked by the right coronary artery, which exited the model there.
2. A small triangular-shaped area in the y - z field, located at the bottom of the y^+ side wall, was also masked from viewing. This was caused by the silicone insert that was made to model the aortic root geometry: it fit seamlessly to the

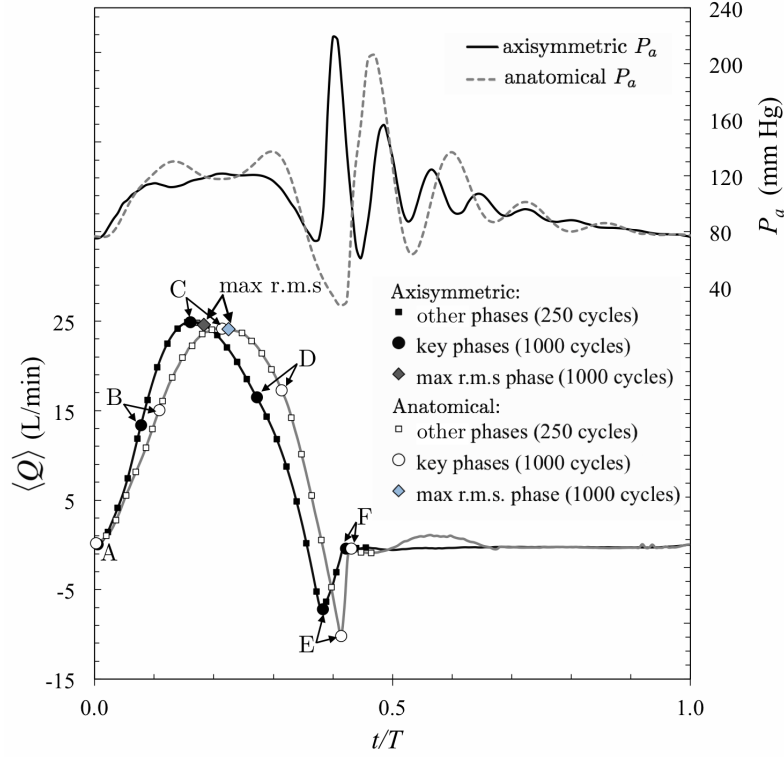


Figure 5.3: Temporal variations of the aortic pressure P_a and the bulk flow rate $\langle Q \rangle$ in the axisymmetric and the anatomical models. The phases for which PIV measurements were made are indicated: “other phases” by small squares; “max-r.m.s.” by diamonds; and “key phases” by large circles labelled: A - start of systole; B - mid-acceleration point; C - peak of the forward flow; D - mid-deceleration point; E - peak reverse flow, and F - valve closure.

inside of the aorta base for most of its perimeter, but protruded slightly into the channel at this masked-out location. It is believed that this protrusion had insignificant effects on the flow patterns.

3. Some imperfections in the anatomical model, such as inhomogeneities in the silicone, caused some streaks to appear in the calculated velocity field. To minimize this streaking, the vectors were calculated using a 24×24 pixel interrogation window size (compared with the 12×12 pixel interrogation window used in the axisymmetric model). Although the results appeared to be meaningful, discretion must be used in their interpretation.

The phase-averaged velocity measured in the aorta model at phase C is shown in Figure 5.4 for the y - z centre-plane and the x - z centre-plane. The images are

plane	region	phases	no. cycles	resolution (μm)
x - z , $y/D = 0$	RT	$t/T = 0.003 : 0.0167 : 0.436$	250	270
	RT	A, B, C, D, E, F	1000	270
	ASC	$t/T = 0.003 : 0.0167 : 0.436$	250	270
	ARCH	$t/T = 0.003 : 0.0167 : 0.436$	250	270
x - z , $y/D = -0.211$	RT	$t/T = 0.225$	1000	270
y - z , $x/D = 0$	RT	$t/T = 0.003 : 0.0167 : 0.436$	250	270
	RT	B, C, D, E	1000	270
	ARCH	$t/T = 0.003 : 0.0167 : 0.436$	1000	270
y - z , $x/D = -0.421$	RT	$t/T = 0.225$	1000	270

Table 5.2: Summary of PIV measurements made in the anatomical model: planes, regions, phases, number of cycles recorded, and vector resolutions. RT - aortic root; ASC - ascending aorta; ARCH - aortic arch.

composites of several fields of view (three for the x - z plane and two for the y - z plane) which were measured independently and collated in post-processing. The coronary arteries are not shown here because they did not coincide with these centre-planes. Measurements on a plane coinciding with the right coronary artery are presented in Chapter 6.

As expected, the predominant flow characteristics were the same as those in the axisymmetric model: the triple jet dominated the flow field (depicted in the x - z plane) and recirculation vortices formed in the sinuses. Some differences were apparent, however, as described in the following.

Unlike in the axisymmetric model, the triple jets were not symmetric about the z -axis ($x=0$) (Figure 5.4, right). The jet nearest the convex wall of the aorta propagated the furthest of the three, while the one nearest the concave wall jet propagated the shortest distance and impinged on the outer wall of the aorta, just above the sino-tubular junction. This asymmetry was consistent with previous studies of flow in the aorta, which found that velocity profiles in the ascending aorta were skewed toward the convex wall during systole [49, 21]. In Figure 5.5, velocity profiles are plotted

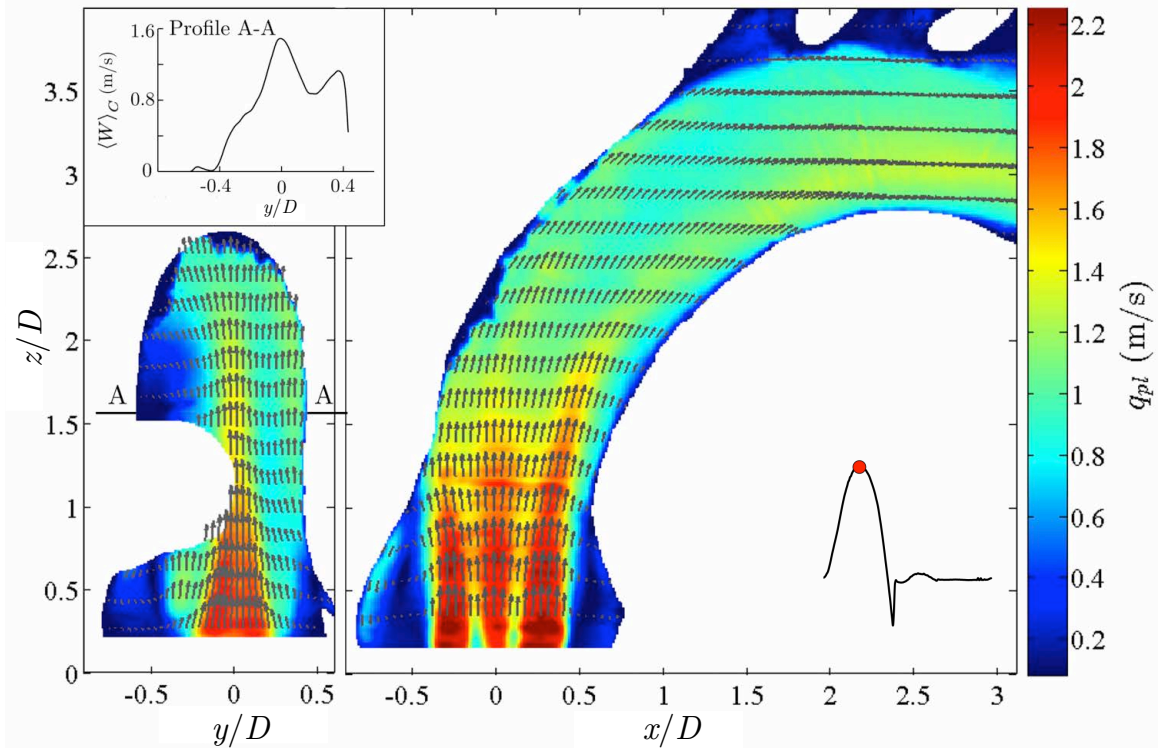


Figure 5.4: Phase-averaged planar velocity magnitude q_{pl} in the anatomical model at peak flow (phase C), for the y - z (left) and x - z (right) centre-planes

for phase C on the x - z planes of both models at various downstream locations. The phase-averaged axial velocity profiles are shown shaded behind the r.m.s. axial velocity fluctuations. At $z/D = 0.35$, the jets do not appear to be skewed and their shapes are nearly identical for the two cases. Jet skewness becomes apparent at $z/D = 0.53$, with higher velocities corresponding to the convex wall. In both geometries, the jets merged as they progressed downstream; this is characterized by the peaks becoming less pronounced, which is shown in Figure 5.5 to occur progressively from $z/D = 0.35$ to $z/D = 1.00$.

The sinuses of the anatomical model curved more gradually than those of the axisymmetric model and resulted in a tear-drop shaped recirculation zone, as opposed to the elongated, ellipse-shaped ones in the axisymmetric model. This is illustrated by Figure 5.6, in which approximate streamlines of the phase-averaged velocity are sketched for the recirculation in both sinus geometries.

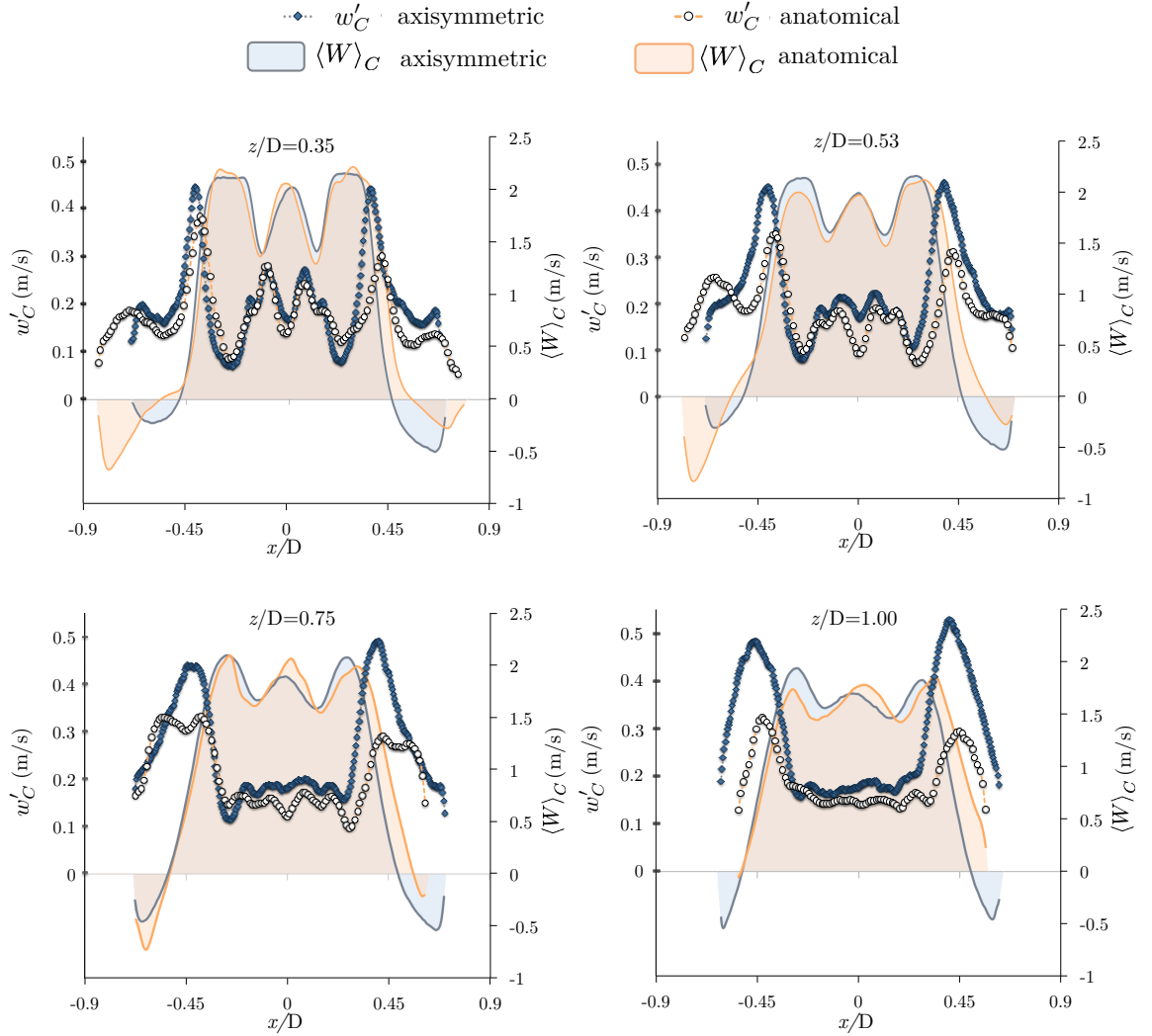


Figure 5.5: Phase-averaged axial velocity profiles $\langle W \rangle_C$ (shaded) and axial velocity fluctuations w'_C (markers) on the x - z centre-planes of the anatomical and axisymmetric models, at peak flow (phase C), for downstream locations $z/D = 0.35$, $z/D = 0.53$, $z/D = 0.75$, and $z/D = 1.00$.

The reverse flow in the right coronary sinus of the anatomical model was stronger than the reverse flows in the other two sinuses of the anatomical model and in the sinuses of the axisymmetric model. This seems to have been caused in part by the deflection of the x^- jet which impinged on the aorta wall, above the sinus. This effect, visible in the x - z velocity field of Figure 5.4, is further illustrated by the velocity profiles in Figure 5.5. Here, it is shown that, compared with the axisymmetric case,

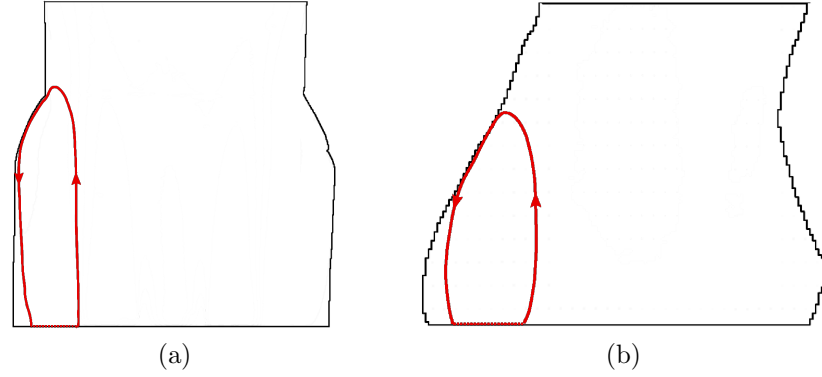


Figure 5.6: Sketch of approximate streamlines of axisymmetric (a) and anatomical (b) right-coronary sinus recirculation at peak flow, in the x - z centre-plane.

the reverse flow in the anatomical model was greater in the right-coronary sinus (the x^- side), but is less in the left-coronary sinus (the x^+ side). This characteristic persisted to $z/D = 0.75$, near the end of the anatomical sinus. In the present valve orientation, the left and right coronary sinuses were approximately symmetric about the valve's plane of symmetry; the effect of valve orientation on the flows in the sinuses and the coronary arteries will be discussed in Chapter 6. It is noted that some differences between the flows measured in the the left and right sinus may be caused by a slight asymmetry of the sinus positions relative to the x - z centre-plane. Although this plane was anterior to the apices of both the left and right sinuses, the left coronary ostium was located further from the plane than the right coronary ostium (Figure 5.1).

Near the valve, at $z/D = 0.35$, the profiles of r.m.s. axial velocity fluctuations w'_C were very similar for the two models (Figure 5.5). The main difference in this region occurred at the outer edges of the lateral jets, where the fluctuations were slightly less in the anatomical model than in the axisymmetric one. This difference grew downstream from the valve: the fluctuations in the axisymmetric model increased to peaks of approximately 0.5 m/s at $z/D = 1.00$, whereas in the anatomical model, the peaks steadily decreased to approximately 0.3 m/s at this point. The fluctuations located between the jets were very similar for the two geometries, but, further downstream of the valve, they were slightly lower in the anatomical model than in the axisymmetric one.

The sinus geometry on the y - z plane was strongly asymmetric about the z -axis: the non-coronary sinus was located in the y^- side of the root, but there was no sinus opposite it. As a result, a large recirculation vortex formed in the non-coronary sinus and a much smaller recirculation vortex on the other side. This effect was propagated up into the ascending aorta, where a low velocity region existed next to the y^- wall, but not at the y^+ wall. In fact, a second jet formed at the y^+ wall, having evolved from the y^+ shoulder, resulting in a double-peaked velocity profile. This is illustrated in the inset of Figure 5.4, which shows the axial velocity profile at $z/D = 1.55$.

Figure 5.7 compares velocity profiles on the y - z planes of the two models for various downstream locations at peak flow. The peak jet velocities were very similar; however, at $z/D = 0.35$ and 0.53 the jet profile was narrower in the anatomical model, as the jet was confined by the vortex in the non-coronary sinus on the y^- side and the aorta wall on the y^+ side. The jet narrowing occurred below the jet ‘shoulder’, and the jet ‘heads’ were nearly identical for the two models. The central jet ‘dimple’, although present in the anatomical case, was not as deep as it is in the axisymmetric model and was not distributed as symmetrically. The velocity profiles reveal that there was virtually no recirculation at the y^+ wall.

The velocity fluctuations on the y - z planes for the two models were remarkably close to each other (Figure 5.7). At the centre location, corresponding to the ‘dimple’, the fluctuations were lower in the anatomical model than in the axisymmetric one, but they were nearly identical at the sides of the jet head. At $z/D = 0.35$, the fluctuations in the anatomical model were less than in the axisymmetric one near both the y^+ and the y^- walls, but at $z/D = 0.53$, they were marginally higher in the non-coronary sinus. This change is consistent with the increase in retrograde velocity in the non-coronary sinus from $z/D = 0.35$ to $z/D = 0.53$ (the peak velocity magnitude increased from 0.20 m/s to 0.40 m/s).

As the flow decelerated, a low-velocity pocket developed at the convex wall of the aortic arch. This is illustrated in Figure 5.8, which shows the planar streamlines of the phase-averaged velocity field at times $t/T = 0.347$ and $t/T = 0.364$ during deceleration; velocity profiles transecting the pocket centre are displayed in the insets. At $t/T = 0.347$, the low-velocity pocket was very large, but the flow was still in the

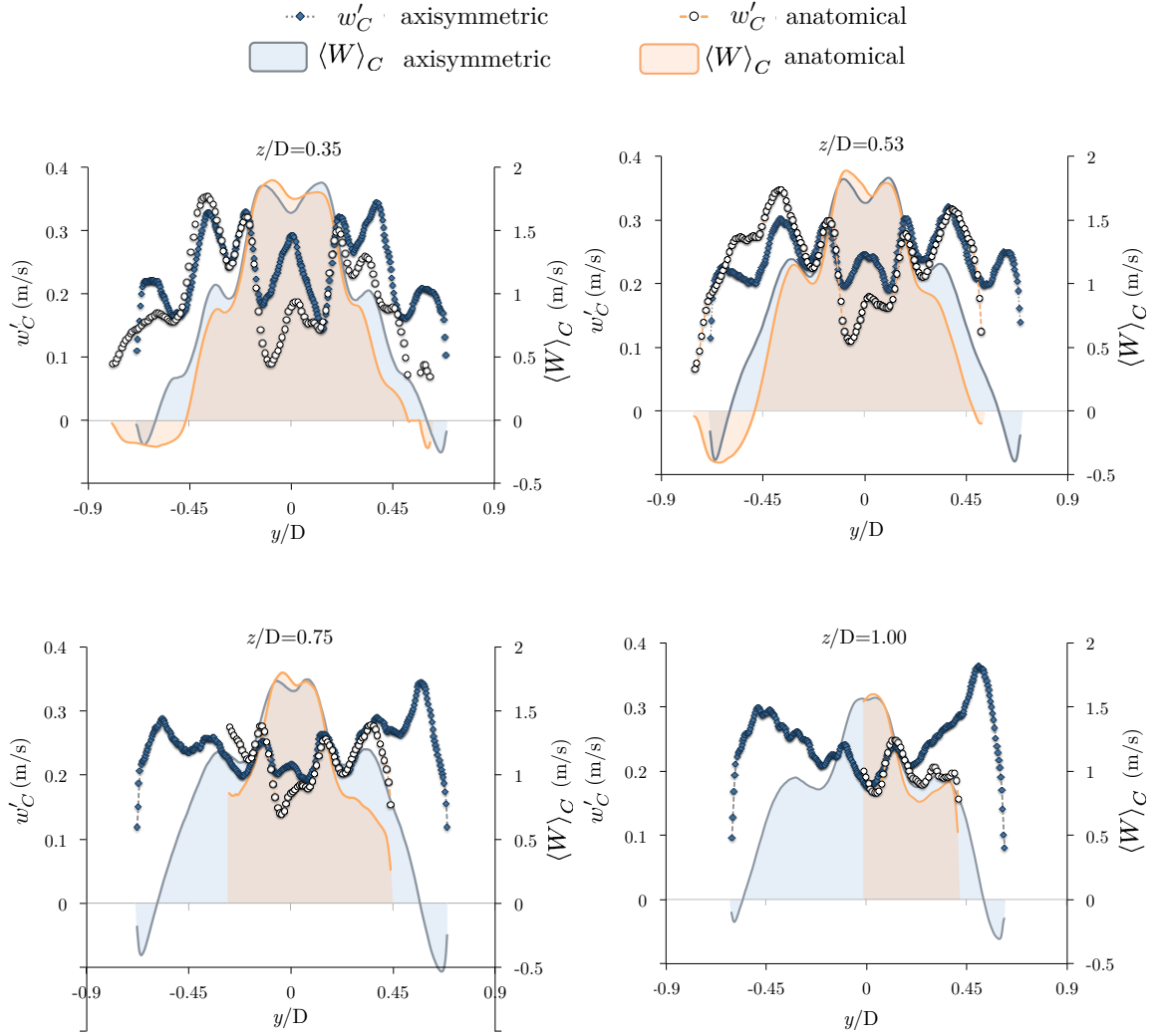


Figure 5.7: Phase-averaged axial velocity profiles $\langle W \rangle_C$ (shaded) and axial velocity fluctuations w'_C (markers) on the y - z centre-planes of the anatomical and axisymmetric models, at peak flow (phase C), for downstream locations $z/D = 0.35$, $z/D = 0.53$, $z/D = 0.75$, and $z/D = 1.00$. (Note that the profiles for $z/D = 0.75$ and $z/D = 1.00$ are incomplete due to the masked area at the coronary artery outlet.)

forward direction across the entire channel. The streamlines of the faster moving fluid bent around the pocket, and the velocity profile was skewed throughout the entire arch, with higher velocity toward the concave wall. The tails of the lateral jets were visible in the velocity profile at section B-B. At phase $t/T = 0.364$, reverse flow occurred at the convex wall. The low-velocity pocket was larger than at previous

phases and extended upstream to the vortex in the left-coronary sinus. The lateral jets were weaker than at $t/T = 0.347$, and their tails were less distinguishable in the velocity profile at section B-B.

5.3 Turbulent stress

The turbulent stress parameter RS_{max} is displayed in Figure 5.9 for peak flow on the x - z and y - z centre-planes of the anatomical and axisymmetric models.

In the x - z centre-plane, the stresses at the channel core and very near the valve were similar for the two geometries, but at distances further downstream, particularly at the outer edges of the lateral jets, large differences occurred. In the axisymmetric model the maximum turbulent stress was located near the x^+ side of the sinotubular junction and exceeded 160 Pa. In the anatomical model, the stress at the corresponding location was less than half of that value, and nowhere, other than directly off the leaflet tips, did the stress exceed 75% of it. Additionally, the regions of elevated turbulent stress were much more narrow in the anatomical model. These descriptions are consistent with those of the axial velocity fluctuations, which were shown in Figure 5.5. The general patterns of turbulent stress distribution were similar for the two geometries, with one exception being a “ λ -shaped” region of moderately high RS_{max} which formed at the location of jet impingement at the aorta wall, above the right coronary sinus in the anatomical model. Despite jet impingement, the levels of turbulent stress here were less than in the corresponding location in the axisymmetric model.

In the y - z centre-planes, the turbulent stress was similar in distribution and magnitude for the two models. One relatively small difference occurred at the y^+ side of the sinotubular junction, where the stresses were slightly lower in the anatomical model. This may be a result of the anatomical sinus geometry contracting more gradually than the axisymmetric sinus. In the axisymmetric model, regions of moderate turbulent stress existed between the central jet and the sinus recirculation, on both sides. In the anatomical model, on the y^- side of the jet, the stress in this region was higher and more widespread compared with the axisymmetric case, likely due to the

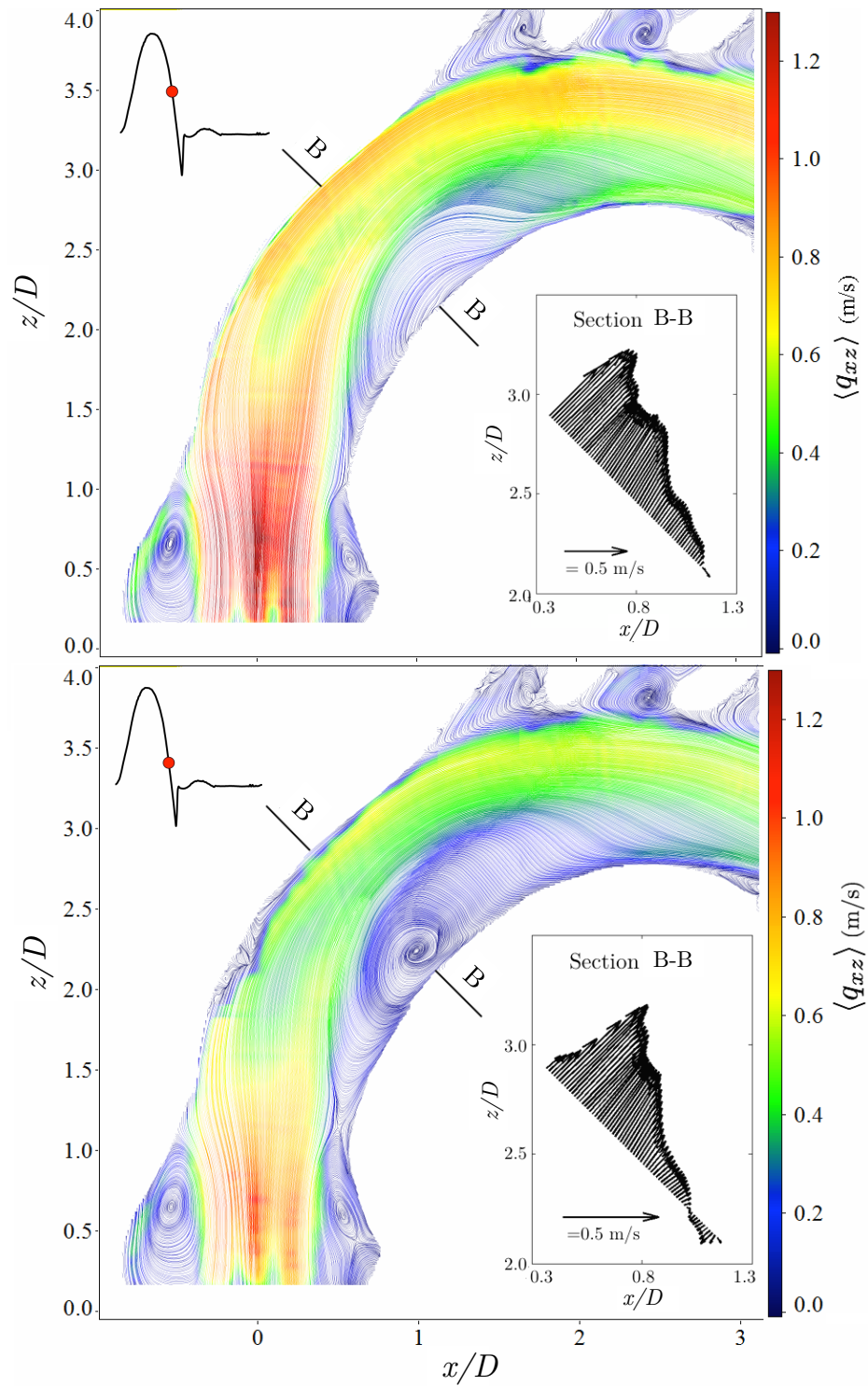


Figure 5.8: Planar streamlines of phase-averaged velocity on the x - z centre-plane of the anatomical model during deceleration times $t/T = 0.347$ (top) and $t/T = 0.364$ (bottom). Insets show the velocity profiles across section B-B, which transects the low-velocity pocket.

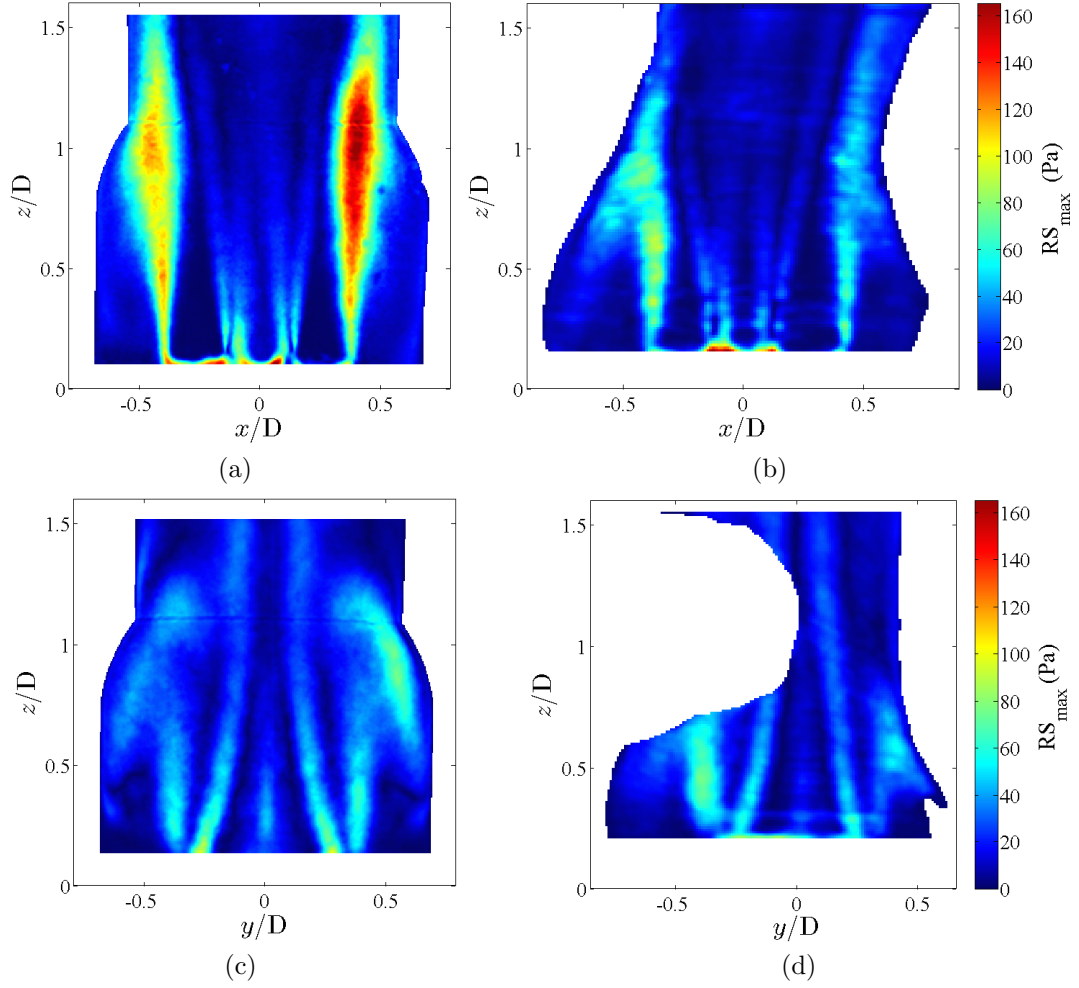


Figure 5.9: RS_{max} for the axisymmetric model (left) and anatomical model (right) at peak flow for the $x-z$ centre-plane (top) and $y-z$ centre-plane (bottom)

stronger recirculation in the non-coronary sinus. On the y^+ side, however, the stress was lower, and this region was indistinguishable from the one of moderate stress at the sinotubular junction.

5.4 Viscous shear stress

For the anatomical model, the viscous shear stress metric τ_{max} was calculated from vectors obtained using a 24×24 interrogation window size and was then adjusted with the corrective mapping algorithm presented in Appendix E.2. The results for the $x-z$ and $y-z$ centre-planes are shown with the corresponding results in the axisymmetric

model in Figure 5.10. Some error is apparent in the resulting maps: there remains some streaking resulting from model imperfections discussed earlier. Additionally, the corrective mapping algorithm is inherently imperfect. Regardless, some general observations and inferences may be made from the result.

In both the x - z and y - z planes, the general patterns of viscous shear stress distribution were similar for the two models. Regions of high viscous shear stress were located at regions of high spatial velocity gradient, most notably at the outsides of the lateral jets. For both models, the viscous shear stress was greatest in the x - z plane, immediately downstream of the valve; however, the stress at this location was highest in the axisymmetric model. The wake of high viscous shear stress at the outsides of the lateral jets was narrower, did not extend as far, and was lower in magnitude in the anatomical model compared with the axisymmetric one. The low stress regions corresponding to the tops of the lateral jets were not as low in the anatomical model. This may be a result of greater jet diffusion (and, consequently, diffusion of shear stress), but it may also be attributed to error caused by the corrective mapping algorithm. Like for the turbulent stress, a “ λ -shaped” region of moderate viscous shear stress existed at the location of jet impingement above the right sinus.

In the y - z plane, the overall viscous shear stress was highest in the anatomical model with the regions of moderate shear stress being more wide-spread than in the axisymmetric model. Like for the turbulent stress, a region of moderately high viscous stress existed between the central jet and the recirculation of the non-coronary sinus. The viscous stress in the corresponding region of the axisymmetric model was lower and more diffuse.

5.5 Analysis of the results and discussion: Anatomical model *vs.* axisymmetric channel

It was found that, in general, flow characteristics immediately downstream of the valve and near the valve axis were very similar for the two models. This was expected, as regions in close proximity to the valve would naturally be influenced more strongly by the valve’s geometry and kinematics than the more distant channel geometry.

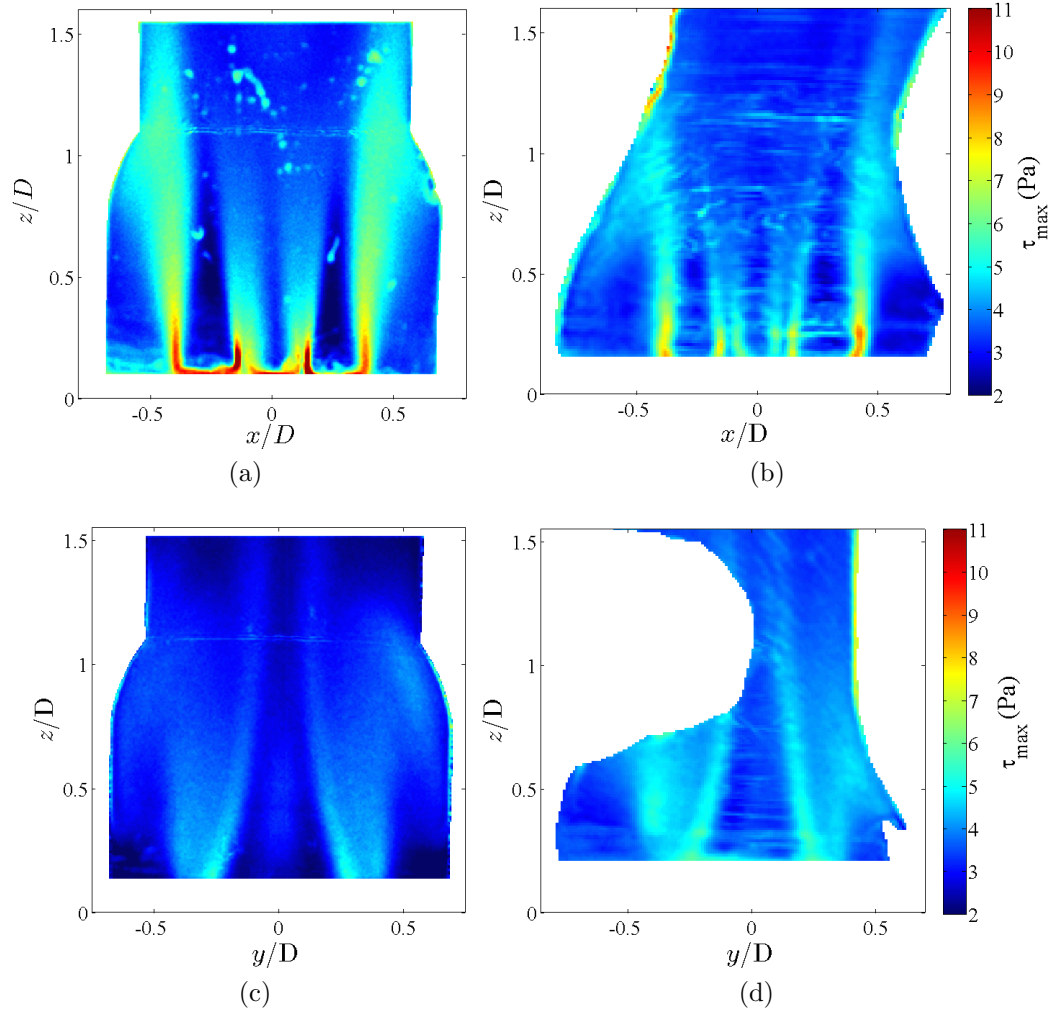


Figure 5.10: τ_{max} for the axisymmetric model (left) and anatomical model (right) at peak flow for the x - z centre-plane (top) and y - z centre-plane (bottom).

Large differences of flow characteristics, however, emerged further downstream and at radial distances away from the z -axis. The most significant differences occurred at the outsides of the lateral jets, where both the turbulent stress and the viscous shear stress were found to be much lower in the anatomical than the axisymmetric model. In view of the uncertainty arising from differences in the processing methods for the two models, the difference in viscous shear stress cannot be decisively attributed to the geometrical differences of the channels. For the turbulent stresses, however, because the uncertainty was found to be relatively small (Chapter 3.4) and conditions of measurement and processing were effectively the same, these differences may be

largely attributed to geometrical differences in the two channels.

5.5.1 The effect of sinus shape

Figure 5.11 shows sample instantaneous planar “streamlines” for the two sinus geometries at several phases during flow acceleration, each together with a planar “streamline” constructed from the phase-averaged velocity map. At mid-acceleration (Phase B), in both geometries, a vortex formed at the base of the sinus between the lateral jet and the channel wall. The vortices were coherent with smoothly shaped boundaries and maintained their shapes from one cycle to the next, indicating that the flow was laminar. As the fluid continued to accelerate, the vortices were advected downstream, stretching towards the end of the sinus. In the axisymmetric model, the vortex was stretched and broke into numerous smaller vortices, thus also generating strong turbulence. In the anatomical model, however, the vortex remained coherent, even at peak flow.

Two characteristics of the sinus geometry are believed to influence the evolution of the vortex: (1) the height to width ratio of the sinus and (2) the angle of the sinus walls. The height to width ratio of the axisymmetric sinus was very large, causing intense stretching of the vortex, which essentially pulled it apart. In contrast, the height to width ratio of the anatomical sinus was much smaller (closer to 1), and the vortex was able to withstand the lesser degree of stretching. The phase-averaged streamlines in Figure 5.11 illustrate the paths of recirculation for the two sinus geometries. At peak flow in the axisymmetric sinus, the fluid was forced to reverse direction within a narrow space and align with the axial direction for remainder of the sinus. In contrast, the anatomical sinus had sloped walls, which permitted the fluid to turn more gradually: at the top of the sinus, the flow was forced to rotate by only about 135° , and at the bottom of the sinus, where the fluid completed its turning, the turning radius was relatively large. Because the turning radius is inversely proportional to vorticity, it is inferred that the vorticity magnitude was less in the anatomical sinus than in the axisymmetric sinus.

Figure 5.5 shows that the velocity gradient component $\frac{\partial \langle W \rangle}{\partial x}$ was slightly lower

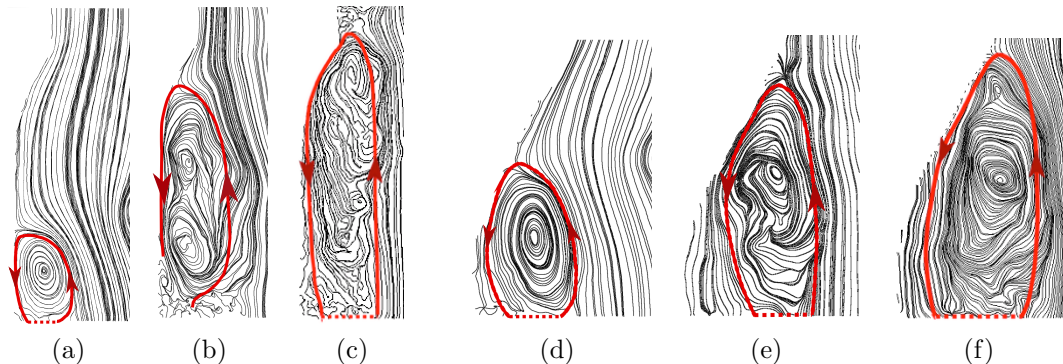


Figure 5.11: Representative instantaneous planar “streamlines” in the axisymmetric sinus (a-c) and the anatomical sinus (d-f) at phase B (a, d), phase C (c, f) and an intermediate phase (b, e) (at a phase difference from phase B that was 0.7 times the difference between phases B and C).

within the sinuses of the anatomical model than in the axisymmetric model, presumably as a result of the larger radius of sinus recirculation. The observed lower level of turbulence in the anatomical model may thus be attributed to lower production by mean shear as well as the limited vortex breakdown. Figure 5.12 illustrates the projection of the velocity on the x - z plane in the x^- sinus at $z/D = 0.75$ for the two geometries at peak flow. It shows that in the anatomical sinus the fluid began to turn into the recirculation flow much closer to the channel axis than in the axisymmetric sinus, in which the flow was closely aligned with the axial direction until very near the centre of recirculation. This illustrates the more gradual turning and the greater transverse spread of the lateral jets in the anatomical sinus. The centre of phase-averaged recirculation was further upstream in the anatomical sinus than in the axisymmetric one; note that the flow from the anatomical profile turned from right to left, indicating that the centre of recirculation was located below the shown profile, while the flow in the axisymmetric model (at the same downstream location) turned from left to right, indicating that the centre of recirculation was above the profile.

The lateral jets in the anatomical model reattached further upstream than in the axisymmetric model (at $z/D \approx 0.97$ compared to $z/D \approx 1.26$ at peak flow). This was caused in part by the increased transverse spread of the lateral jets and the larger

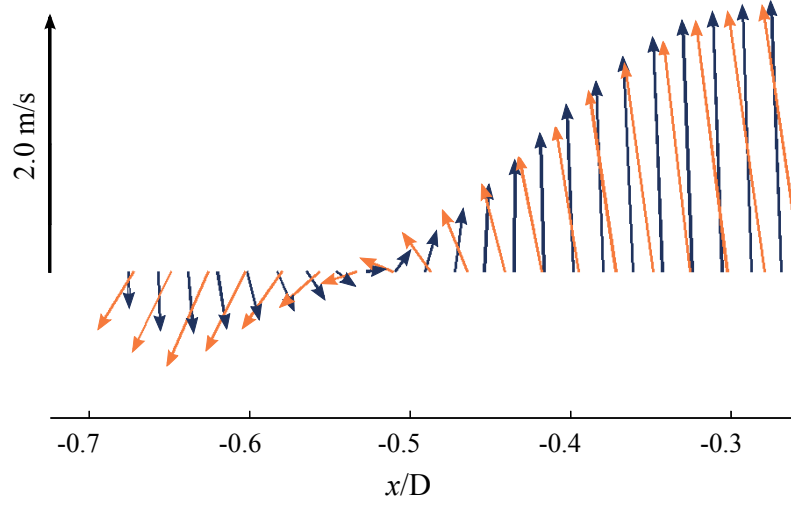


Figure 5.12: Phase-averaged velocity vector projections on the x - z plane in the x^- sinus of the axisymmetric (dark blue arrows) and anatomical models (orange arrows) at $z/D = 0.75$.

turning radius in the anatomical sinus. This coincides with the finding of Borazjani *et al.* [12] that the jet reattached more upstream in an anatomical geometry than in an axisymmetric one. Earlier jet reattachment has the effect of reducing the length of the recirculation region and, consequently, reduces the length of the jet-recirculation shear layer, in which high levels of turbulent and viscous stresses are present.

In addition to the sinus geometry, other factors may have contributed to the transverse spread of the lateral jets. For example, Borazjani *et al.* [12] suggested that secondary flow structures resulting from the aorta curvature contributed to the transverse spread of the jets by impeding their downstream motion. It is possible that secondary flow structures contribute to the reduction of turbulent and viscous stresses in other ways such as by stabilizing the shear layers between the jets. The measurement of these structures, however, was not in the scope of this study.

In summary, the shape of the sinuses at the base of the aorta was found to have significant effects on the turbulence past the BMHV. In general, it was found that a sinus cavity with a low height-to-width ratio and gradual changes in geometry engendered lower turbulence than a cavity with a high height-to-width ratio and abrupt changes in geometry. The results suggest that the shape of the anatomical

sinus similarly contributed to the lower level of viscous stresses in it. The more gradual and coherent recirculation in the anatomical sinus was accompanied by a relatively mild instantaneous shear and consequently relatively low viscous stresses.

5.5.2 The effects of non-axisymmetry and curvature

The effect of sinus non-axisymmetry was most noticeable in the y - z plane, where the non-coronary sinus occupied the y^- side and no sinus was present on the y^+ side. As a result of this asymmetry, an imbalance developed in which a large region of low velocity flow occupied the non-coronary sinus and the region directly downstream of it in the ascending aorta, while high velocities occurred at the opposite wall. This could be problematic *in vivo* by fostering thrombus growth (as explained in Section 2.3), or by altering the normal shear stress distribution at the adjacent vessel walls, which could promote arterial disease [58]. Positioning the BMHV as far as possible into the non-coronary sinus at implantation may minimize the size of this low velocity region. Also, it is of interest how this region would be affected by valve orientation: e.g. would the region be smaller or would the velocity imbalance be less strong if the valve were rotated? This relationship will be investigated in Chapter 6.

The curvature of the aorta was found to affect the velocity distribution measured in the x - z plane in the following ways.

- As a result of the curvature of the ascending aorta toward the x^+ direction, the x^- side lateral jet impinged on the concave wall at the end of the right-coronary sinus. The deflection of this jet probably caused the increase in retrograde flow observed in the right coronary sinus. This flow asymmetry would tend to boost the flow in the right coronary artery, while abating the flow in the left coronary artery. It is unclear, however, whether such effects, which were observed during systole, would have a significant impact on the overall coronary flow rates, because the majority of coronary flow occurs during diastole. These effects on coronary flow can have great clinical importance: flow impediment in the coronary arteries can result in angina pectoris or cardiac ischemia, conditions marked by chest pain that occur when the heart doesn't get adequate blood

supply [39]. The dependence of jet impingement location and coronary flow rates on valve orientation will be investigated in Chapter 6.

- Stresses associated with the x^- lateral jet were lower toward the concave wall than the convex one. This is illustrated in Figure 5.13, which shows q'_{xz} , the r.m.s. velocity fluctuations of the planar velocity q_{xz} , during deceleration ($t/T = 0.281$). It has been previously found that jet impingement in blood vessels can cause elevated levels of wall shear stress at the site of impingement [7] and that such increases in the stresses acting on the vessel wall could cause morphological changes in the wall tissue [58]. Here, although there was a slight increase in the levels of turbulent and viscous stresses at the site of impingement, these levels were less than at the same location in the axisymmetric model at peak flow (Figure 5.9).
- During acceleration and at peak flow, the velocity was skewed towards the convex wall. On the x^- side, the skewing had the effect of a decreased phase-averaged shear, which could have contributed to the reduced turbulent and viscous stresses on this side. On the x^+ side of the aortic root, the shear was equal to or lower than in the straight geometry despite the skewing towards this side. This may have been due to the increased transverse spread of the jets in the anatomical model.
- During deceleration, the velocity was skewed toward the concave wall, and a low velocity pocket formed at the convex wall of the arch. The stresses associated with the x^+ jet extended far into the arch, and wrapped around the concave side of the low velocity pocket (Figure 5.13). Despite the large expanse of this region, the highest levels of velocity fluctuations in it were less than in the axisymmetric model at a comparable phase of the cycle.

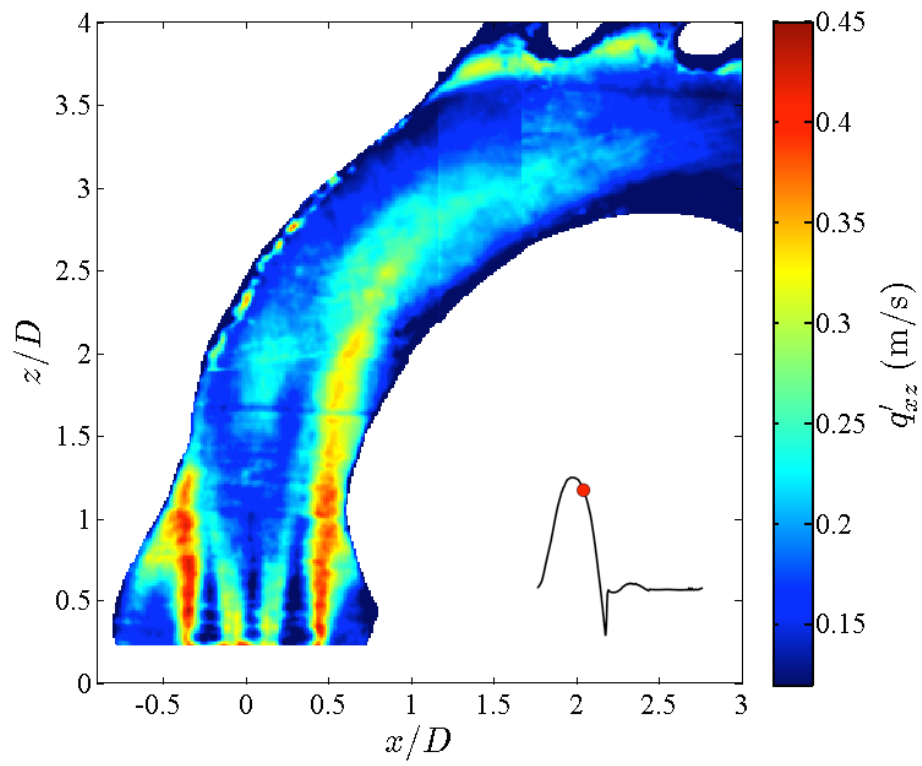


Figure 5.13: The standard deviation of velocity fluctuations q'_{xz} in the x - z centre-plane of the anatomical model at $t/T = 0.281$ during deceleration.

Chapter 6

Results: Velocity Measurements for Three Valve Orientations in an Anatomical Aorta Model

6.1 Measurement locations and sampling conditions

Measurements were made with the BMHV mounted at the inlet of the anatomical aorta model in one of three orientations (Figure 6.1):

1. with the valve’s plane of symmetry aligned normal to the plane of curvature, called the “normal” orientation (the same orientation as for Chapter 5);
2. with the valve’s plane of symmetry aligned parallel to the plane of curvature, called the “aligned” orientation;
3. with the valve’s plane of symmetry rotated 22 degrees (counter-clockwise looking upstream) from the plane of curvature, called the “inclined” orientation; in this orientation, the valve’s plane of symmetry intersected approximately the middle of the right coronary ostium.

Measurements were first made for each valve orientation on the x - z and y - z centre-planes, with all of the branching arteries clamped, as done previously. Then, measurements were made at an off-centre x - z plane that intersected approximately the middle of the opening to the right coronary artery ($y/D = 0.165$). The clamp at the exit of this artery was fully opened, allowing fluid to flow through it. Measurements of the right coronary flow were made on that plane for each of the three valve orientations. The measurement locations and sampling conditions for the different valve orientations are listed in Table 6.1.

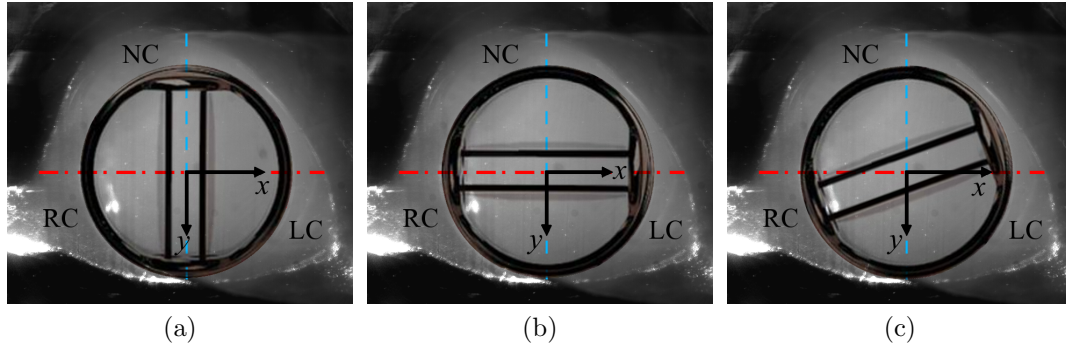


Figure 6.1: Valve orientations with respect to plane of aorta curvature: (a) “normal”; (b) “aligned”; (c) “inclined”. RC - right-coronary sinus; LC - left-coronary sinus; NC - non-coronary sinus; red dashed line - x - z centre-plane; blue dashed line - y - z centre-plane.

6.2 Centre-plane measurements

6.2.1 Planar velocity maps

The phase-averaged velocity maps measured on the x - z centre-plane at phases B and C are shown in Figure 6.2 for the normal, aligned and inclined valve orientations. For the normal valve orientation, the x - z plane was perpendicular to the valve’s symmetry plane and transected the three jets. These results were described previously in Section 5.2.

For the aligned valve orientation, the x - z plane was parallel with the valve symmetry plane and, like the y - z plane for the normal orientation, revealed the profile of the central jet. At phase B, the jet was seen to emerge past the valve leaflets into the aortic root. The velocity in the sinuses was very low - lower than that for the normal orientation at this phase. At phase C, the profile of the central jet was very narrow. Its axis was aligned with the z -axis upto approximately $z/D = 1.25$, after which point it curved slightly in the direction of the aorta curvature. At this location, a second velocity peak, lower in magnitude than the peak of the central jet, formed near the concave wall. This was similar to the double peaked profile that developed past the valve in the y - z plane for the normal orientation (Figure 5.4). As for the normal orientation, the reverse flow in the right coronary sinus was stronger than

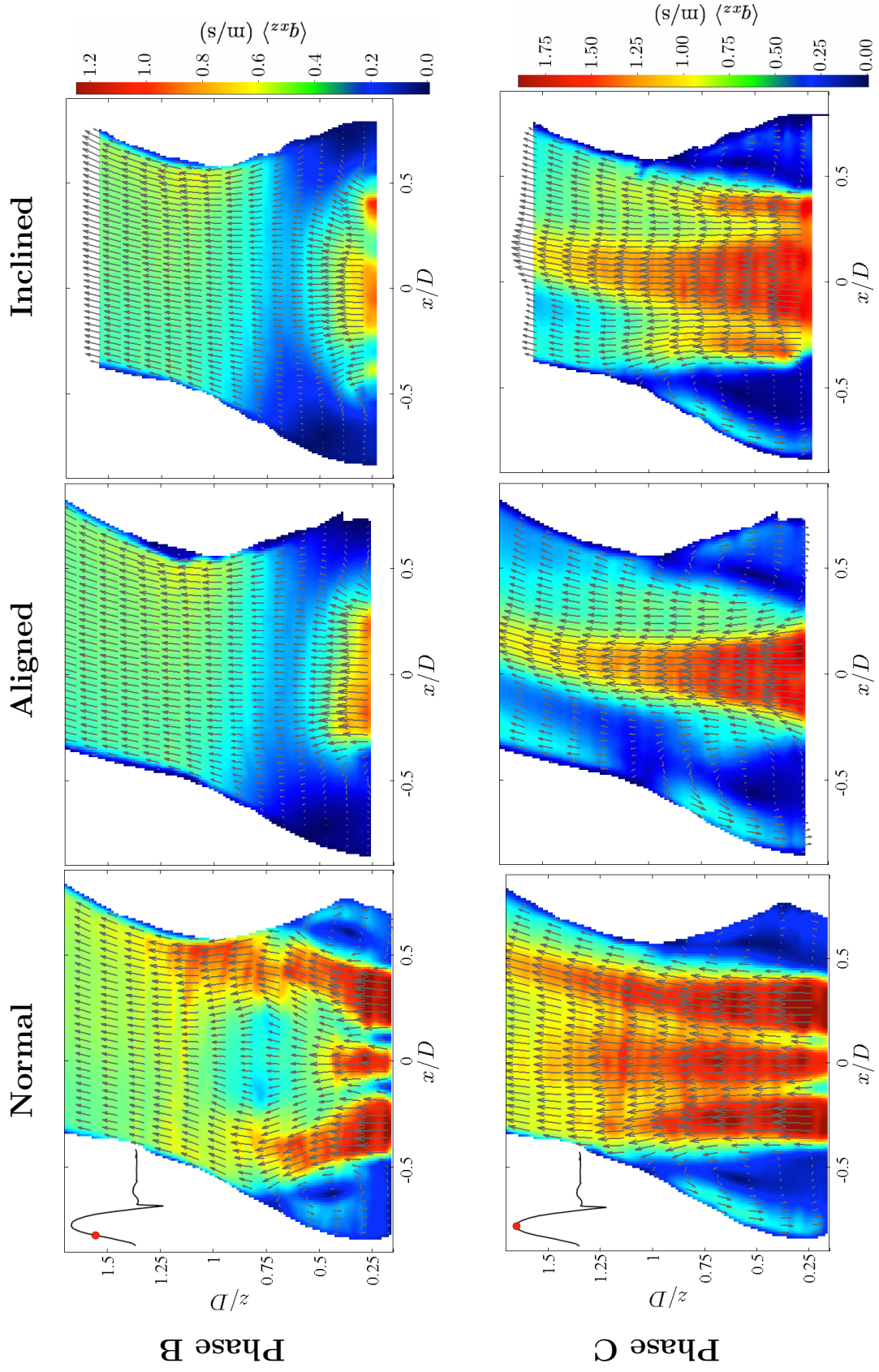


Figure 6.2: Phase-averaged velocity q_{xz} on the x - z centre-plane at phases B (top) and C (bottom), for valve orientations: normal (left column), aligned (middle column) and inclined (right column).

plane	valve ori- entation	flow condition	phases	no. cycles
x - z , $y/D = 0$	normal	no flow	$t/T = 0.003 : 0.0167 : 0.436$ B, C, D	250 1000
x - z , $y/D = 0$	aligned	no flow	$t/T = 0.003 : 0.0167 : 0.436$ B, C, D	250 1000
x - z , $y/D = 0$	inclined	no flow	$t/T = 0.003 : 0.0167 : 0.436$ B, C, D	250 1000
y - z , $x/D = 0$	normal	no flow	$t/T = 0.003 : 0.0167 : 0.436$ B, C, D	250 1000
y - z , $x/D = 0$	aligned	no flow	$t/T = 0.003 : 0.0167 : 0.436$ B, C, D	250 1000
y - z , $x/D = 0$	inclined	no flow	$t/T = 0.003 : 0.0167 : 0.436$ B, C, D	250 1000
x - z , $y/D = 0.165$	normal	RC flow	$t/T = 0.003 : 0.0167 : 0.436$ B, C, D	250 1000
x - z , $y/D = 0.165$	aligned	RC flow	$t/T = 0.003 : 0.0167 : 0.436$ B, C, D	250 1000
x - z , $y/D = 0.165$	inclined	RC flow	$t/T = 0.003 : 0.0167 : 0.436$ B, C, D	250 1000

Table 6.1: Summary of PIV measurements made in the anatomical model for varying valve orientation and flow in right coronary artery: planes, phases, and number of cycles recorded.

that in the left coronary sinus.

For the inclined valve orientation, the x - z centre-plane was oblique to the valve symmetry plane and passed at an angle through the valve's centre orifice and through the corners of the valve's lateral orifices, as illustrated in Figure 6.3. As for the aligned orientation, the velocities in both sinuses at phase B were near zero. At phase C, as for the other two orientations, the reverse velocity was stronger in the right sinus than in the left. The left sinus recirculation, however, was weaker than in the aligned

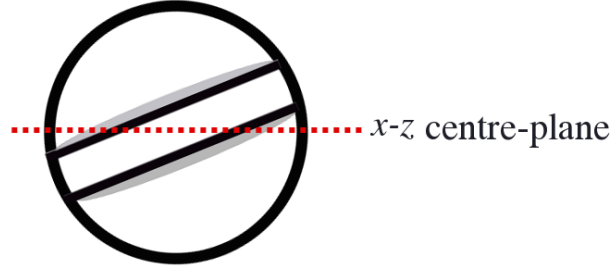


Figure 6.3: Top-down view of the BMHV in the inclined orientation, relative to the x - z centre-plane.

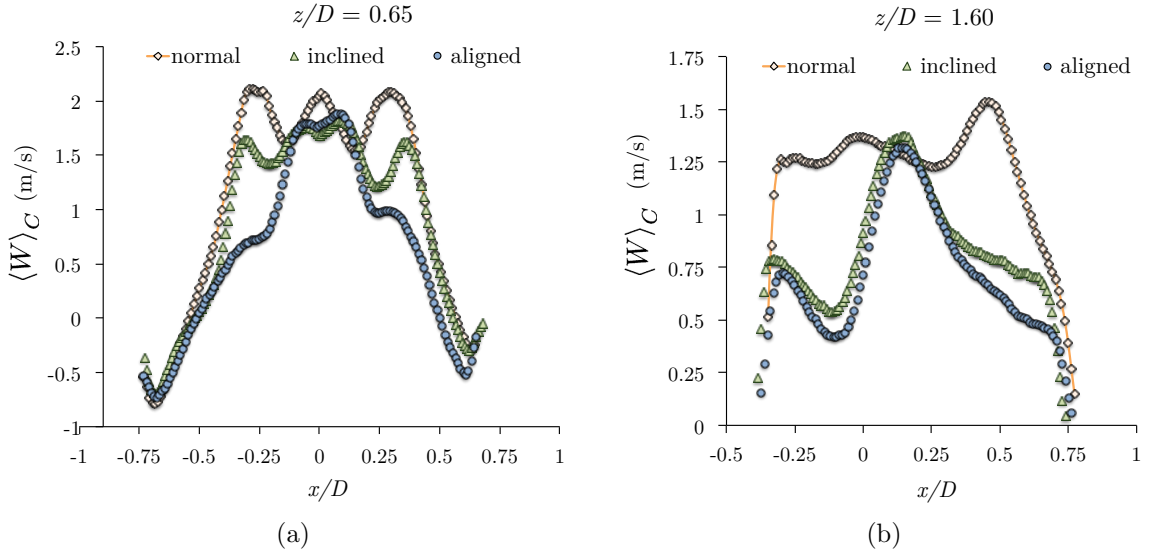


Figure 6.4: Axial velocity profiles on the x - z centre-plane at $z/D = 0.65$ and 1.60 for normal, inclined and aligned valve orientations at peak flow.

valve orientation, but similar to that for the normal orientation. The projection of the central jet extended much further downstream than the projections of the lateral jets on this plane.

Figure 6.4 shows the axial velocity profiles $\langle W \rangle$ in the x -direction for the three valve orientations at $z/D = 0.65$ and 1.60 . The profiles at $z/D = 0.65$, which was approximately two-thirds down the length of the sinus, reveal that the reverse velocities in the right coronary sinus (on the x^- side) were nearly the same for the three orientations, but the reverse velocity in the left sinus was higher for the aligned than for the normal and inclined orientations. Further downstream, at $z/D = 1.60$,

the triple jets of the normal orientation were less pronounced than at $z/D = 0.65$, as they merged and spread; they were also slightly skewed toward the x^- side. The profile for the aligned orientation at $z/D = 1.60$ shows that a secondary velocity peak developed at the x^- wall. On the x - z plane, the average velocity was higher for the normal orientation, in which the plane transected the three jets, compared to the aligned orientation, in which the plane was parallel to the valve symmetry plane. Unsurprisingly, the averaged profile velocity for the inclined orientation was intermediate between those for the other two cases.

The phase-averaged velocity measured on the y - z centre-plane at phases B and C are shown in Figure 6.5 for the three valve orientations. For the valve in the normal orientation, the y - z plane illustrates the profile of the central jet. These results were previously discussed in Section 5.2.

For the aligned valve orientation, the y - z plane illustrates, a cross-section of the triple velocity jets. At phase B, the lateral jet on the y^+ side appears to have deflected off of the channel wall, which similarly occurred at phase B for the normal valve orientation (as shown by the x - z plane in Figure 6.2). At phase C, the three jets were aligned with and symmetric about the z -axis. The recirculating flow in the non-coronary sinus was even slower than for the normal valve orientation at this phase.

For the inclined orientation, the y - z plane shows oblique cross-sections of the three jets, which were similar to those shown for the aligned orientation on this plane. In contrast to the velocity maps on the x - z plane for this orientation, in which the central jet was relatively wide and the lateral jets were very narrow, the jet profiles on the y - z plane were all comparable in width. At phase B, the lateral jet on the y^- side was stronger than the one on the y^+ side. Like in the aligned orientation, the y^+ side jet appears to have deflected off of the y^+ side wall of the channel. At phase C, the jet profiles were narrower than for the aligned orientation, and their velocities were slightly lower. This was expected because the y - z plane did not pass through the centres of the lateral jets (where the jets were strongest and widest) for the inclined orientation, but it did for the aligned orientation. The velocities in the non-coronary

sinus were comparable to those for the aligned orientation, but lower than those for the normal orientation.

6.2.2 Turbulent stress

The maximum turbulent stress RS_{max} is shown in Figure 6.6 for the three valve orientations at peak flow on the x - z and y - z centre-planes. As was found in Section 5.3, regions of high turbulent stress occurred at the outsides of the lateral jets and, to a lesser degree, between the lateral jets and the central jet (shown in planes perpendicular to the valve symmetry plane) and at the sides of the central jet (shown in planes parallel to the valve symmetry plane).

The levels of turbulent stress at the outsides of the lateral jets were generally higher for the normal configuration compared with the aligned configuration; the corresponding peak value was 10% higher (92 Pa compared to 84 Pa). Also, the regions of high stress were slightly larger for the normal orientation; e.g., for the aligned orientation, the high stress region at the outside edge of the y^- side lateral jet was much more narrow than the corresponding region of the x^- side jet for the normal orientation. The turbulent stresses between the lateral jets and the central jet, however, were generally higher for the aligned compared to the normal orientation; peak values were 12% higher (56 Pa compared with 50 Pa), and the regions of high turbulent stress between the jets were more expansive.

The levels of turbulent stress on the planes parallel to the valve symmetry plane (the y - z and the x - z planes for the normal and aligned orientations, respectively) were generally higher for the normal orientation. The average stress on the y - z plane for the normal orientation was 46% higher than on the x - z plane for the aligned orientation (19 Pa and 13 Pa, respectively), and the peak stress was 9% higher (83 compared to 76 Pa). The stresses at the sides of the central jet were higher for the normal than the aligned orientation, and the regions of high stress that for the normal orientation were (1) between the central jet and the non-coronary sinus recirculation, and (2) at the y^+ side wall had no comparable regions on the x - z plane for the aligned orientation. One region of significant turbulent stress for the aligned orientation actually formed

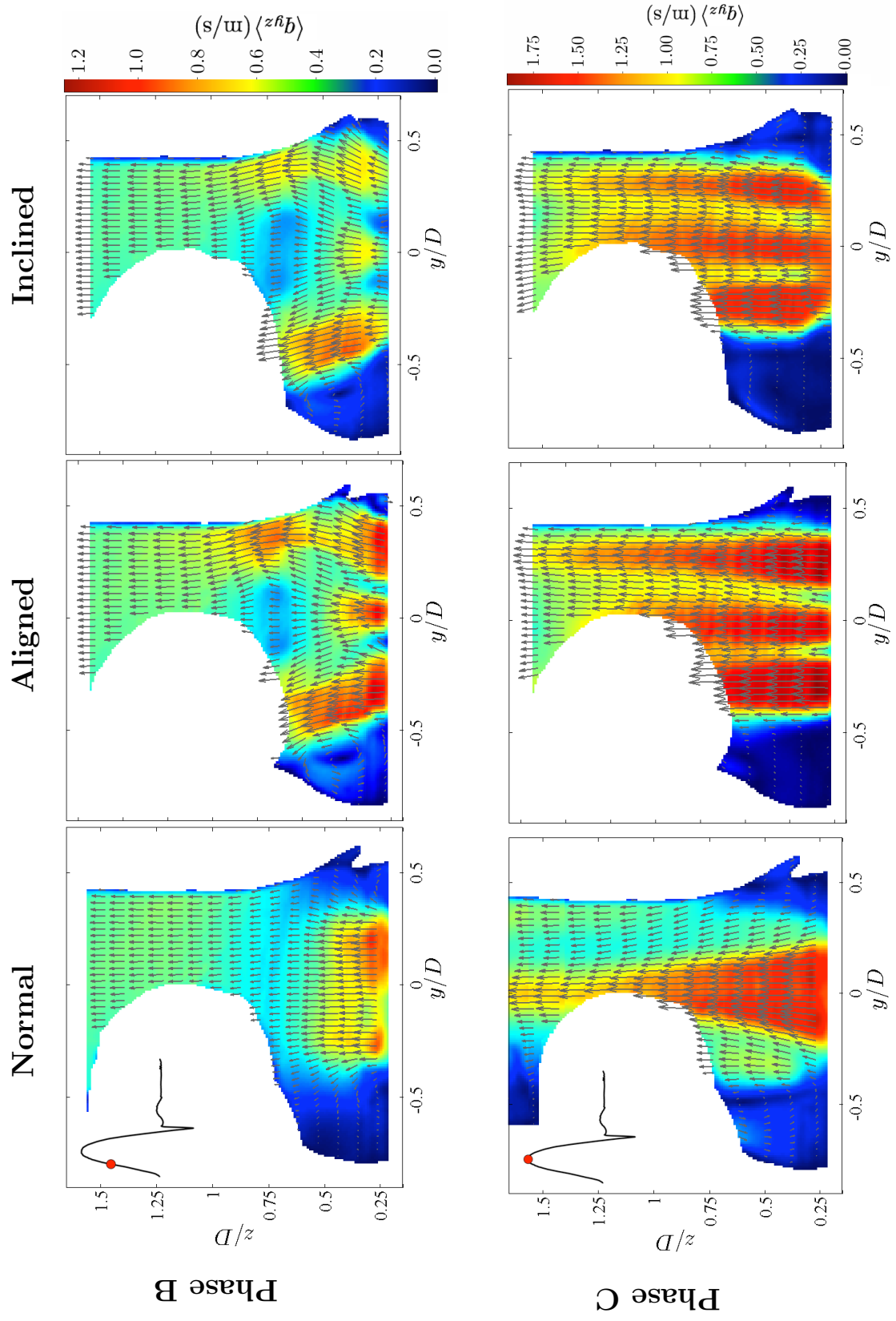


Figure 6.5: Phase averaged velocity q_{yz} on the y - z centre-plane at phase B (top), and C (bottom) for valve orientations: normal (left column), aligned (middle column) and inclined (right column).

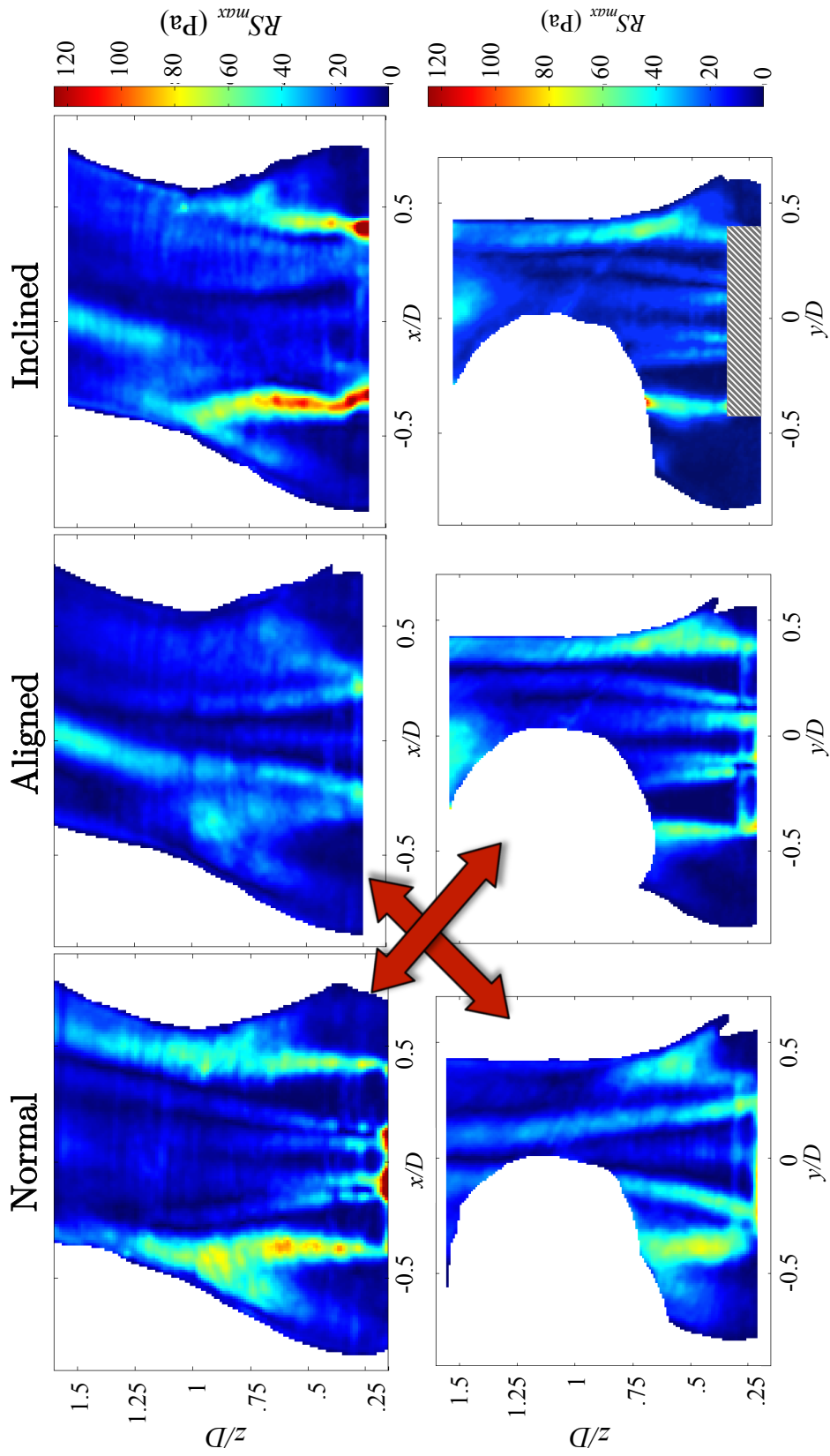


Figure 6.6: RS_{max} at peak flow for valve orientations: normal (left), aligned (middle), and inclined (right), in the x - z centre-plane (top) and the y - z centre-plane (bottom). Red arrows connect the planes that were perpendicular to the valve's symmetry plane, and the planes that were parallel to it. The region for the inclined orientation marked with cross-hatching was believed to be strongly affected by error, attributed to loss of optical access, and was not considered in the analysis.

near the side of the central jet towards the concave wall, at $z/D \approx 1.5$, which was the approximate location where the jet began to curve with the aorta curvature.

Considering the measurements from both the x - z and y - z planes, the average value of RS_{max} was 21% higher for the normal valve orientation than the aligned one (18.5 Pa compared to 15.3 Pa).¹

For the inclined valve orientation, two streaks of high turbulent stress levels are shown on the x - z plane. These streaks corresponded to the flow emerging from the corners of the valve's lateral orifices. The turbulent stress corresponding to these regions was even higher than at the outsides of the lateral jets for the normal orientation (peak values of 205 Pa compared to 92 Pa). The average stress on this plane was 6% higher for the inclined orientation than the normal orientation. The streaks of high shear stress extended to the end of the sinuses. The flow emitted from the orifice corners would have been subject to sharp transitions and separation from the valve housing and from the leaflets, which were very close together on this plane.

The region at the base of the root just past the valve on the y - z plane for the inclined orientation was believed to be strongly affected by error, attributed to loss of optical access, and therefore will not be considered in the analysis. The y - z plane reveals a turbulent stress distribution similar to that measured for the aligned orientation on the same plane: regions of high stress were present at the outsides of the lateral jets and, to a lesser extent, between them and the central jet. The stresses were slightly lower in value, and the regions of moderate stress were less extensive than for the aligned orientation, for which the plane bisected the three valve orifices. Omitting the masked region at the base of the root, the average and peak stresses on the y - z plane for the inclined orientation were 16% and 25% less than those for the aligned orientation, respectively.

6.2.3 Viscous shear stress

Figure 6.7 shows the phase-averaged viscous shear stress τ_{max} at peak flow for the three valve orientations on the x - z and y - z centre-planes. Normal to the valve's

¹The average was calculated over the flow domain within the ranges $z/D = 0.2$ to 1.5 , $x/D = -0.8$ to 0.83 (x - z planes) and $y/D = -0.85$ to 0.7 (y - z planes).

plane of symmetry (the x - z plane for the normal orientation and the y - z plane for the aligned orientation), high levels of viscous shear stress existed at the shear layers of the jets; such shear layers formed between the lateral jets and the valve housing, between the lateral jets and the valve leaflets, and between the central jet and the leaflets.

From these images, on the planes perpendicular to the valve's symmetry plane, the shear stress levels at the base of the shear layers were somewhat higher for the normal orientation than for the aligned orientation: the maximum averaged τ_{max} for the normal orientation was 7% higher than for the aligned orientation (8.8 Pa *vs.* 8.2 Pa). For the planes parallel to the valve's symmetry plane, maps of the phase-averaged shear stress indicate that the stress was also higher for the normal orientation compared to the aligned orientation (on the y - z and x - z centre-planes, respectively).

To show the shear stress distribution for each valve orientation, histograms were generated from the instantaneous measurements, as was done in Section 5.4. The measurements from both the x - z and y - z centre-planes were used to generate the histogram for each orientation, which was then normalized to obtain the PDF $f(\tau_{max})$. The PDFs for the normal and aligned valve orientations, shown in Figure 6.8, confirm the observations made from the maps of phase-averaged τ_{max} . The measurements for the normal valve orientation had slightly more instances of high level shear stress than for the aligned valve orientation: the percentage of measurements that exceeded 10 Pa on the x - z and y - z planes were 1.7% and 1.3% for the normal and aligned valve orientations, respectively.

A PDF was not made for the inclined valve orientation, because, unlike for the other two orientations, the measurement planes weren't normal and parallel to the valve symmetry plane, thereby making inappropriate the comparison of the resulting PFD to the others.

From the maps of phase-averaged τ_{max} for the inclined orientation (Figure 6.7), streaks of high viscous shear stress were shown on the x - z centre-plane, corresponding to the flow emitted from the corners of the lateral orifices. These stresses, however, were not as high as at the outer shear layers of the lateral jets, shown on planes

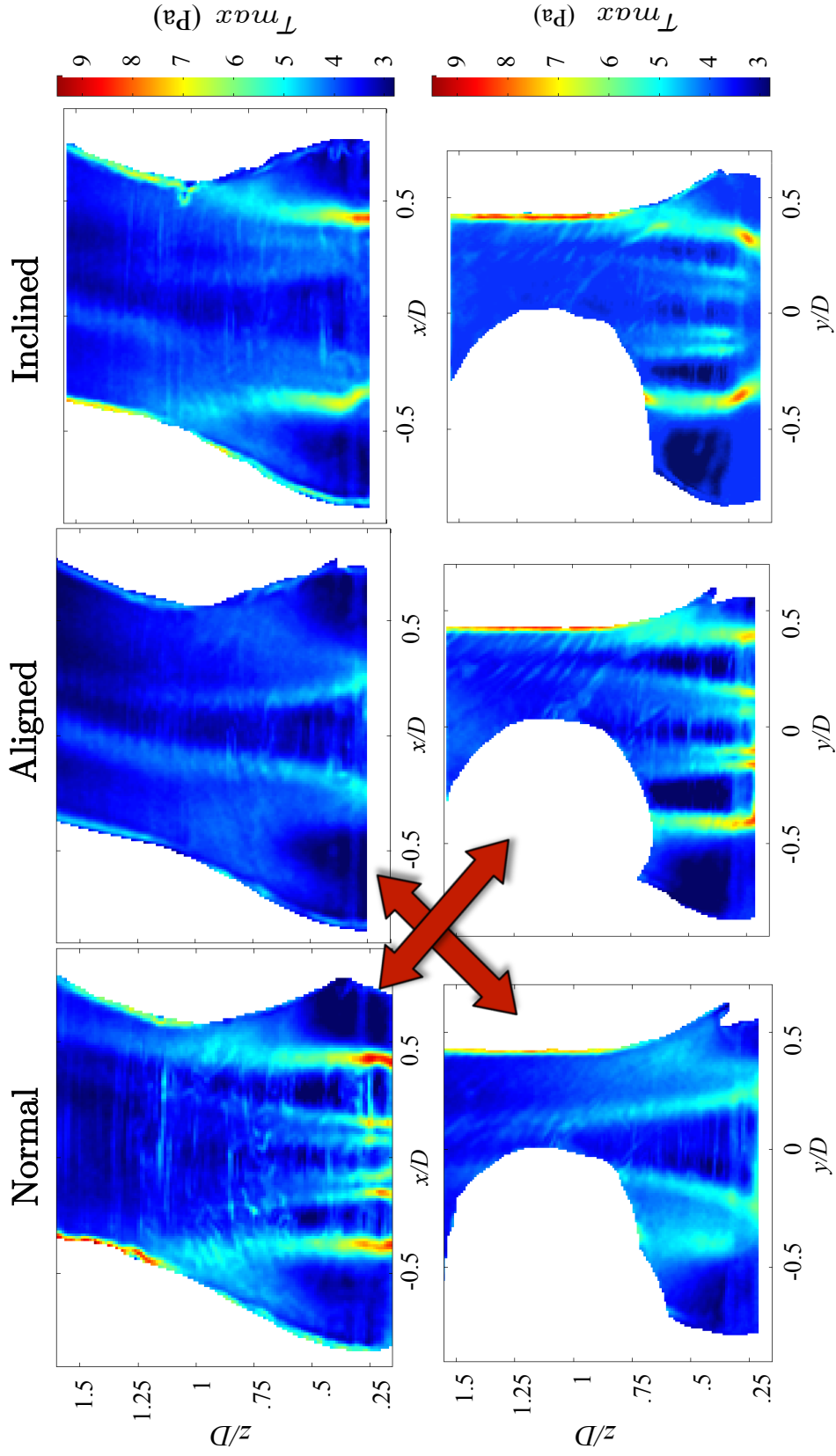


Figure 6.7: Averaged τ_{max} at peak flow for valve orientations: normal (left), aligned (middle) and inclined (right), in the x - z centre-plane (top) and the y - z centre-plane (bottom). Red arrows connect the planes that were perpendicular to the valve's symmetry plane, and the planes that were parallel to it.

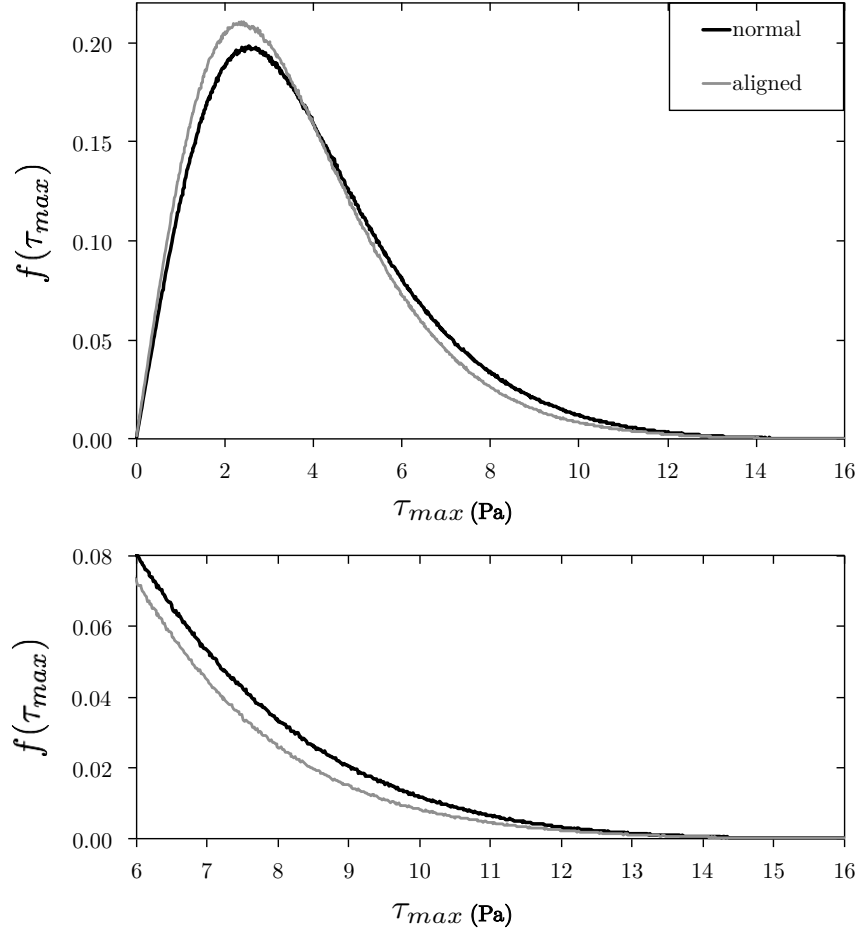


Figure 6.8: Normalized histogram of τ_{max} for normal and aligned valve orientations, on the x - z and the y - z centre-planes.

normal to the valve symmetry plane. For the inclined orientation on the y - z plane, the stress distribution was similar to those on planes perpendicular to the valve's plane of symmetry, with high levels of stress at the jet shear layers; however, the levels of turbulent stress were not as high.

6.3 Measurements of coronary flow rate

Measurements were made on the plane that intersected the middle of the right coronary ostium, with the clamp at the artery outlet fully open, letting fluid flow through the artery. Here, the results are presented for the three valve orientations for various

time instances during the forward flow cycle.

Figure 6.9 shows the phase-averaged planar “streamlines” in the x - z plane at three time instances during flow acceleration and peak flow for the three valve orientations. At $t/T = 0.086$, for all orientations, the flow was in the forward direction and fluid entered the coronary artery. For the normal valve orientation, the x^- lateral jet appeared to be mid-way into the aortic root, whereas for the aligned and inclined orientations, no jets were visible. In the normal orientation, the sinus recirculation vortex formed between the lateral jet and the sinus wall, but it was small and did not seem to influence the coronary flow.

At $t/T = 0.108$, the central jet had extended into the root for the aligned and inclined orientations, and the sinus vortex formed between it and the sinus wall. The vortex was larger for the inclined than the aligned orientation, extending further towards the channel centre. The velocity in the coronary artery was higher than at the previous phase for the aligned and inclined orientations, and of all orientations, was highest in the aligned one. For the normal orientation, the sinus vortex was further downstream than previously and extended past the opening of the coronary artery. Flow from the lateral jet wrapped around it to enter the artery.

At $t/T = 0.214$ (peak flow), for all orientations, the sinus vortex occupied the entire sinus region and blocked flow from entering the coronary artery. A second vortex is shown inside the artery, indicating that the flow recirculated.

Figure 6.10 shows the planar streamlines for the three valve orientations at three time instances during flow deceleration. Slightly after peak flow, at $t/T = 0.261$, the flow in the aorta had begun to decelerate and, for all orientations, the sinus vortices retreated slightly upstream, partially unblocking the coronary ostium, so that flow again wrapped around the vortex and entered the artery. As the fluid was diverted around the vortex, it entered the artery at an angle to the artery’s axis. As a result, in all orientations, the flow separated at the downstream-side entrance to the artery, and the forward coronary flow was confined to the upstream side of the vessel. The area of separation was smaller for the inclined valve orientation, where the flow entered the vessel in a direction more aligned with its axis. Additionally, the sinus vortex occupied a larger region of the sinus for the inclined orientation than for the others,

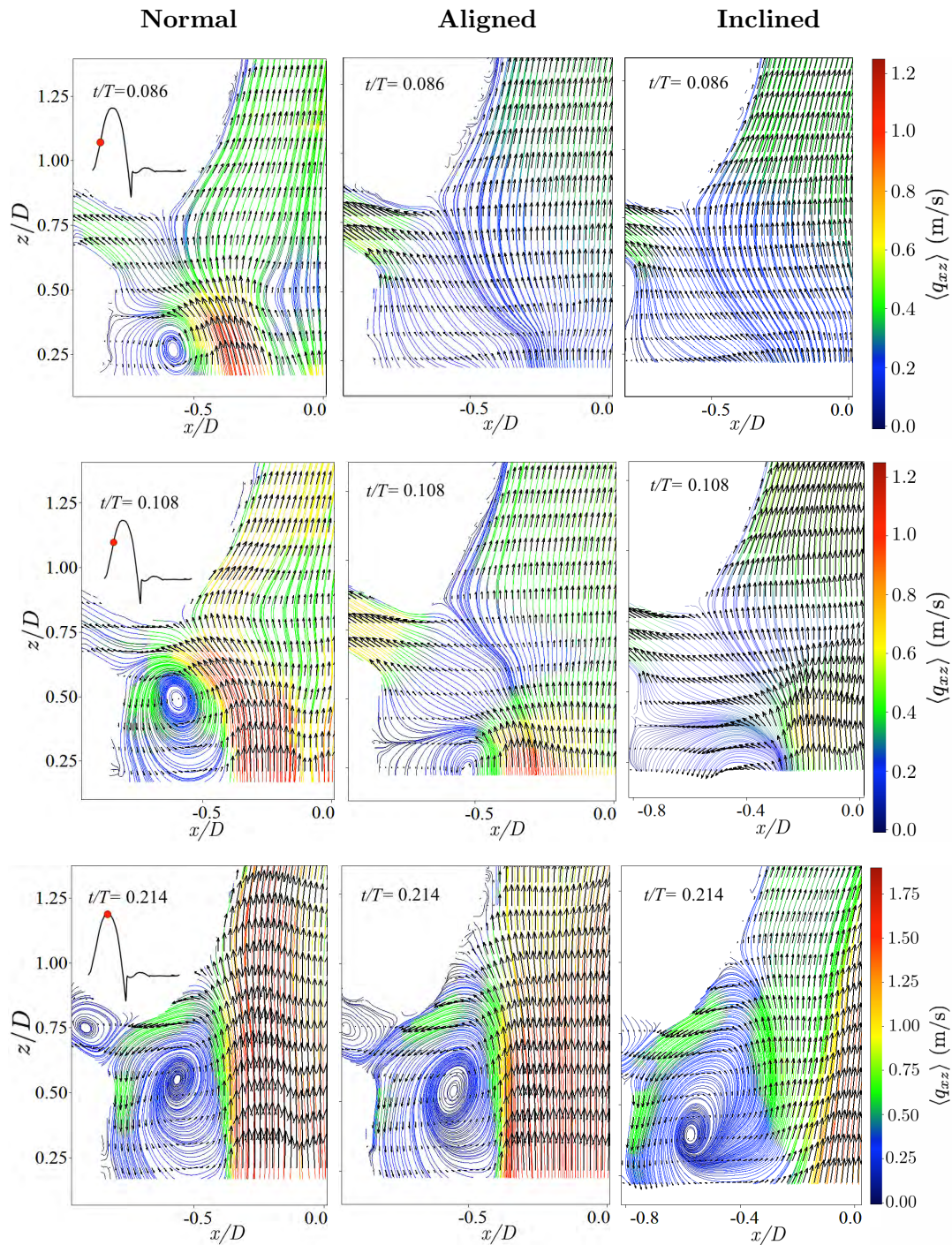


Figure 6.9: Phase-averaged planar “streamlines” of flow in the right coronary artery for accelerating phases $t/T = 0.086$ (top row) and $t/T = 0.108$ (middle row), and peak flow ($t/T = 0.214$) (bottom row) for normal (left column), aligned (middle column), and inclined (right column) valve orientations.

and included flow nearer the channel centre. The phase-averaged centre of vortex recirculation for this orientation was located further upstream than for the other orientations; it was located the furthest downstream for the normal valve orientation.

At $t/T = 0.314$, for the aligned and inclined orientations, forward flow continued to enter the coronary artery; however, for the normal orientation the artery became blocked again by the sinus vortex causing flow to again cease in the artery.

At $t/T = 0.364$, for all valve orientations, the sinus vortex was located slightly inward toward the channel centre, and reverse flow drained from the artery between the vortex and the sinus wall. Maps of other time instances not shown here revealed that the reverse coronary flow began at an earlier phase for the normal orientation than for the other two. At $t/T = 0.364$, the reverse flow was strong for this orientation, and a second vortex formed between the drainage jet and the sinus wall. The sinus vortex was squeezed between the drainage jet and the lateral jet coming from the BMHV. For the aligned and inclined orientations, at $t/T = 0.364$, the drainage flow was weaker and followed the sinus wall. Maps of later time instances not shown here showed that the drainage flow for these orientations was strengthened and that a second vortex formed between the drainage jet and the sinus wall, like that shown for the normal orientation at $t/T = 0.364$.

The bulk flow rate in the right coronary artery was estimated by integrating the velocity in the direction of the artery axis across the channel, and assuming that the arterial flow was axisymmetric. The procedures used for this calculation are described in the following.

First, the direction of the artery's axis was determined. While viewing the velocity map in MATLAB, two lines were identified marking the artery walls: one each at the upper and lower walls of the channel, shown in Figure 6.11 as "line 1" and "line 2", respectively. The slope of each line was determined and designated m_1 and m_2 , respectively. The slope of the artery axis was estimated as the average of m_1 and m_2 , and the angle between the artery axis and the x -axis was determined as

$$\alpha_A = \tan^{-1}((m_1 + m_2)/2). \quad (6.1)$$

A vector was then made at an angle normal to the channel axis, $\alpha_N = \alpha_A + \frac{\pi}{2}$,

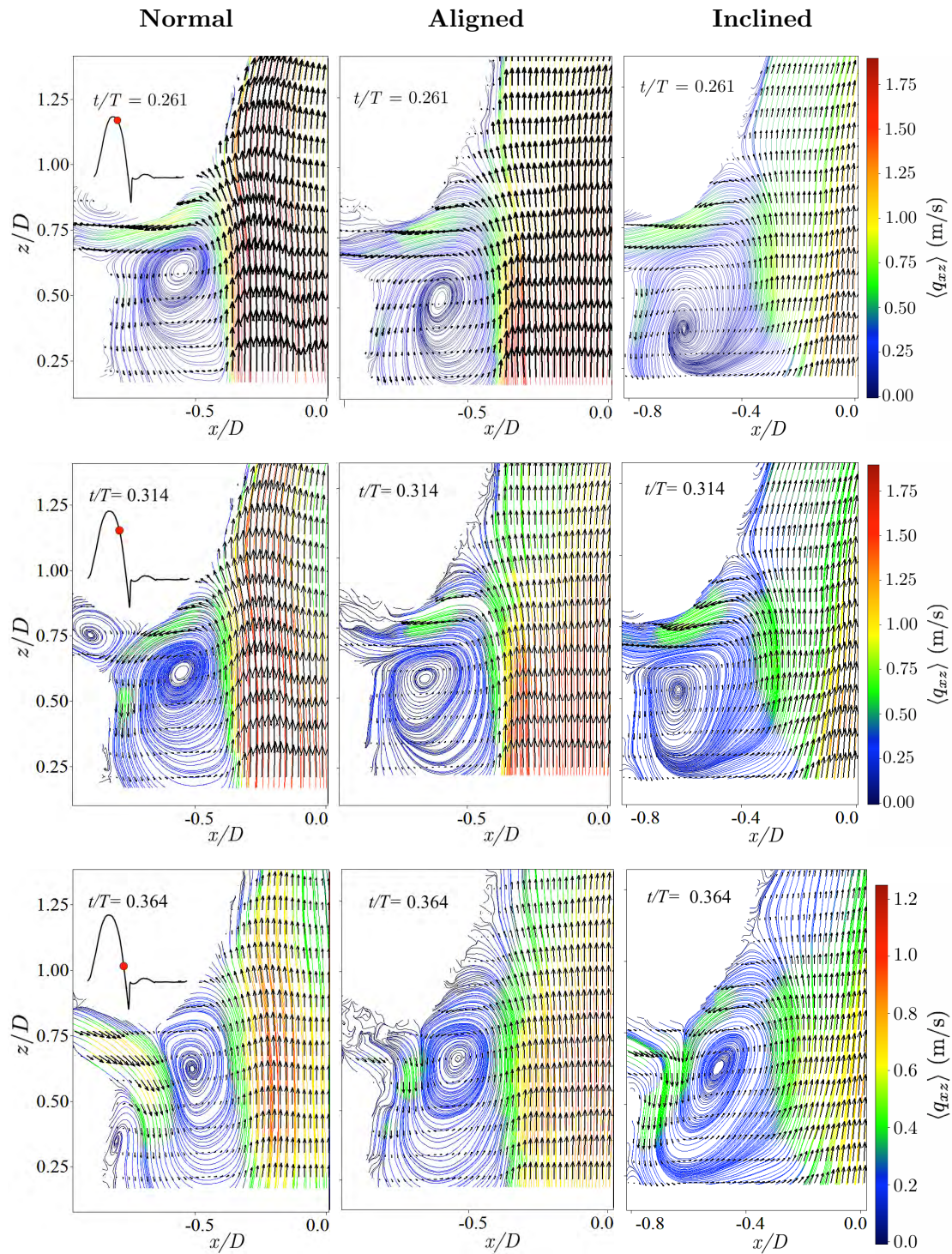


Figure 6.10: Phase-averaged planar "streamlines" of flow in the right coronary artery for decelerating phases $t/T = 0.261$ (top row) and $t/T = 0.314$ (middle row), and $t/T = 0.364$ (bottom row) for normal (left column), aligned (middle column), and inclined (right column) valve orientations.

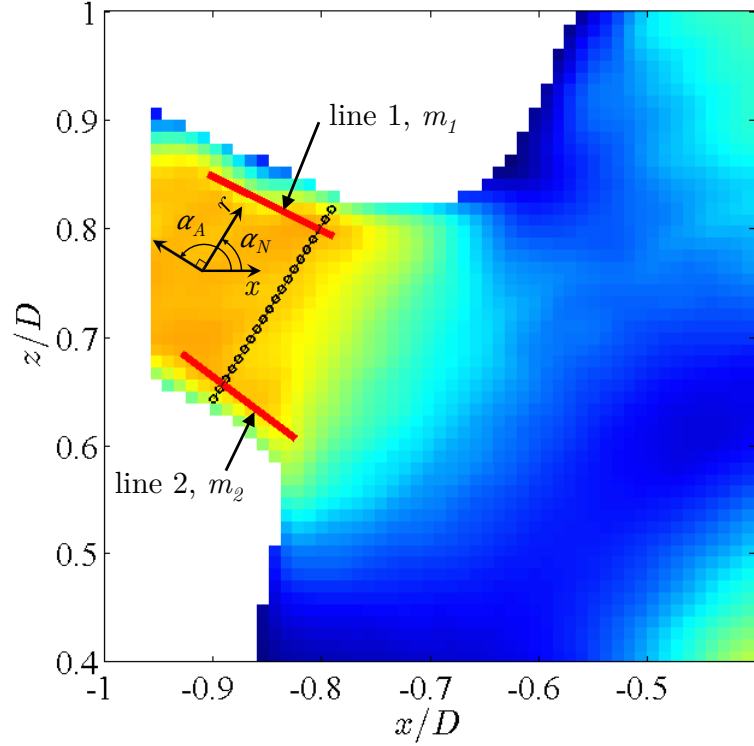


Figure 6.11: Sketch of right coronary artery with lines to determine axial direction. The colouring indicates planar velocity magnitude q_{xz} , the values of which are not essential to the purpose of the sketch.

and discretised across the channel (Figure 6.11). The phase-averaged velocity components $\langle U \rangle$ and $\langle W \rangle$ were determined for each point on the profile by interpolating between the nearest PIV measurements. The velocity in the axial direction, $\langle V_{RC} \rangle$, was determined from these as

$$\langle V_{RC} \rangle = \langle U \rangle \cos(\alpha_A) + \langle W \rangle \sin(\alpha_A). \quad (6.2)$$

As an example, Figure 6.12 shows the axial velocity $\langle V_{RC} \rangle$ vs. the coronary radius r , for the aligned valve orientation at $t/T = 0.086$. Using this result, the bulk coronary flow rate was estimated for each phase as

$$\langle Q \rangle = \pi \int_{r=-R}^{r=R} V_{RC} r dr, \quad (6.3)$$

where $R = 0.103D$ is the radius of the right coronary artery.

The results for the three valve orientations are plotted in Figure 6.13. This shows that the majority of flow occurred at two times during the forward part of the cycle:

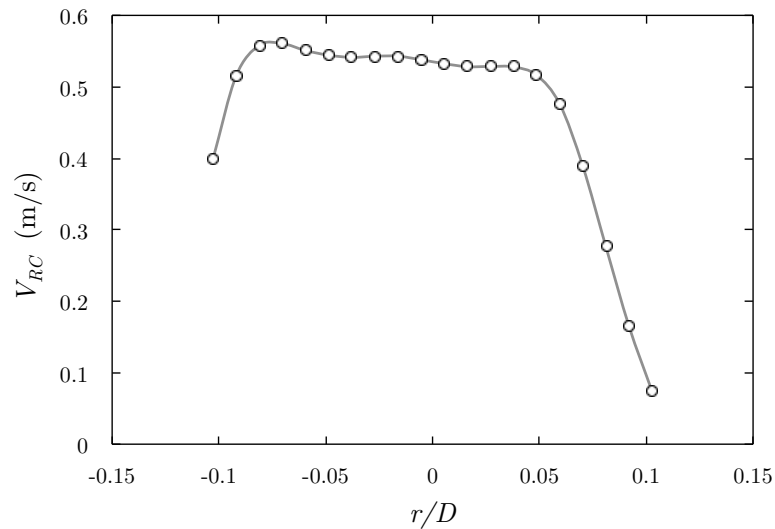


Figure 6.12: Axial velocity V_{RC} in right coronary artery at $t/T = 0.086$ for the aligned valve orientation.

around mid-acceleration and around mid-deceleration. For all three valve orientations, the coronary flow rate was near zero around peak flow due to the blockage of the ostium by the sinus vortex. Positive coronary flow resumed during deceleration, when the sinus vortex retreated slightly upstream, partially unblocking the coronary ostium. The vortex blocked the ostium a second time during late deceleration, again reducing coronary flow, before moving inwards towards the channel, allowing reverse flow to drain from the coronary artery.

During flow acceleration, the coronary flow rate was greatest for the aligned valve orientation and least for the normal orientation. During deceleration, however, the coronary flow was greatest for the inclined orientation and least for the aligned orientation. The volume of fluid that entered the coronary artery was estimated by integrating the flow rate over the duration of forward coronary flow for each orientation. This volume was lowest for the normal orientation (0.74 mL), slightly greater for the aligned orientation (0.84 mL), and greatest for the inclined orientation (1.2 mL).

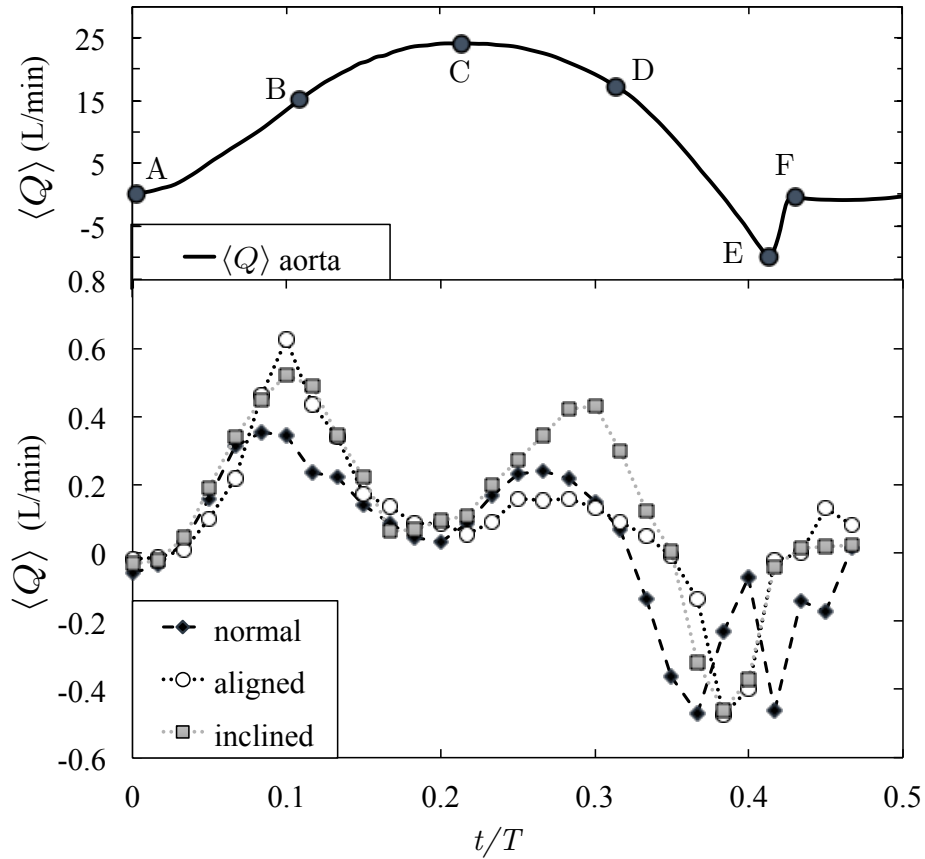


Figure 6.13: Top: flow rate in the aorta during the forward flow phase of the cycle, with key phases indicated by letters: A - start of systole; B - mid-acceleration point; C - peak of the forward flow; D - mid-deceleration point; E - peak reverse flow, and F - valve closure. Bottom: flow rate in the right coronary artery during the forward flow phase of the cycle for normal, aligned and inclined valve orientations.

6.4 Analysis of the results and discussion

6.4.1 Effects of valve orientation on velocity distribution

Measurements for the normal valve orientation in Chapter 5 have shown that the reverse flow in the right coronary sinus was much higher than in the left sinus. This was considered to be the result of the lateral jet impinging on and reflecting off of the artery wall just above the right coronary sinus. Measurements for the aligned and inclined valve orientations, however, showed that, at peak flow, the reverse flow in the right sinus was just as high as it was for the normal orientation, and was similarly much higher than in the left sinus. This finding indicates that lateral jet

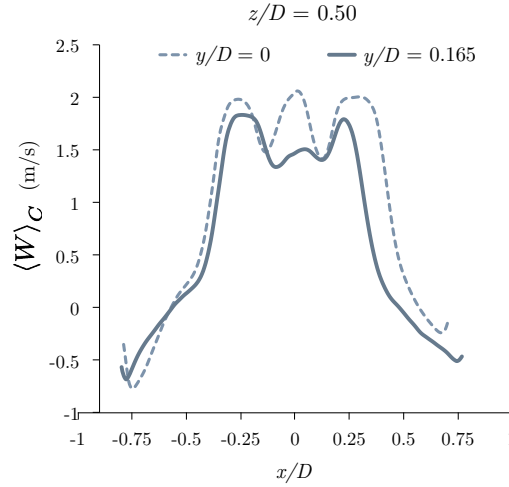


Figure 6.14: Axial velocity profiles for the normal valve orientation in the x -direction.

impingement cannot be the cause of the difference in sinus flows, as there was clearly no jet impingement or reflection for the aligned valve orientation. In the following, we will seek a more plausible explanation for this difference.

Figure 6.14 shows the axial velocity profiles for the normal valve orientation in the x -direction at $z/D = 0.5$ and $y/D = 0$ (centre-plane) and 0.165. These profiles show that the reverse velocity in the left sinus (on the x^+ side) was higher at $y/D = 0.165$ compared to that on the centre-plane (it is noted that the former location was closer to the apex of the left sinus than the latter). This partially explains why, on the x - z centre-plane, the flow was measured to be higher in the right sinus than the left: the reverse flows increased towards the sinus apexes and the centre-plane was closer to the right apex than the left. Even so, this only partially accounts for the difference, as at $y/D = 0.165$ the reverse flow was still slightly greater in the right sinus than the left.

It was also found that, on the x - z centre-plane, the reverse flow in the left coronary sinus was greater for the aligned valve orientation than for the normal and inclined orientations. This may be because the central jet was narrower on this plane for the aligned orientation and left more room for recirculation than the central jets for the two other orientations, which were wider due to their merging with the lateral jets.

It is also noted that, in all valve orientations, one of the lateral jets deflected off

the channel wall at mid-acceleration (phase B). This was because, at this phase, the lateral jets were inclined outward and had a significant lateral velocity component. Although this deflection is noteworthy, it is not likely to have any clinical significance as it occurred at a time in the cycle when the velocities and stress levels were relatively low.

From the measurements on the y - z plane for the normal orientation, it was shown that the low velocity region of the non-coronary sinus extended far into the ascending aorta. For the aligned and inclined orientations, however, the field of view was limited to the aortic root and did not include the ascending aorta. As a result, it was not possible to determine the extent of the low velocity region for these valve orientations. It was interesting to note, however, that the velocity in the non-coronary sinus was even lower for both of these orientations than for the normal one. One may postulate that, because the lateral velocity components of the three jets were very small at peak flow, the low-flow region would extend fairly deeply into the ascending aorta. As was shown in Figures 6.2 and 6.5, however, the fluid velocities near the channel walls were typically higher on planes perpendicular than on planes parallel to the valve's plane of symmetry, due to the spread of the lateral jets. As a result, the low velocity region (measured on the y - z centre-plane) would probably not have extended as far for the aligned and inclined orientations as it did for the normal valve orientation.

6.4.2 Effects of valve orientation on turbulent and viscous shear stresses

The turbulent stresses were on average 21% higher for the normal valve orientation than for the aligned orientation. Additionally, the normal valve orientation resulted in 31% more occurrences of viscous shear stresses that were high enough to potentially activate platelets. These results suggest that the aligned orientation may be preferable over the normal one from the viewpoint of minimizing the stresses acting on blood elements, and reducing the potential for thrombosis.

Conversely, Borazjani *et al.* [12] found that the valve orientation had no significant effect on the viscous shear stress downstream of the valve, but recommended this same aligned orientation because it resulted in the most synchronous valve closure

and lowest leaflet rebound; they attributed this effect to a fairly uniform pressure distribution on the leaflets in the aligned orientation, whereas for other orientations, the two leaflets were unequally affected by the pressure gradient imposed by the aorta curvature. These authors postulated that this configuration would be less susceptible to intermittent regurgitation.

Kleine *et al.* [47] found that, from measurements made in two pigs, the Reynolds normal stresses were lowest when a BMHV was oriented with one lateral orifice toward either the posterior wall or the right-posterior wall of the aorta; these orientations corresponded approximately to the aligned and the inclined orientations of the present study, respectively. This result was fairly consistent with that of the present study.

One possible explanation for the higher stress levels in the normal valve orientation is that the pressure imbalance acting on the leaflets caused by the aorta curvature may have resulted in increased leaflet oscillations and therefore higher velocity fluctuations. From Figure 6.6, however, it is shown that the largest differences in turbulent stress for the two orientations occurred *not* in the shear layers off of the leaflet tips, but in the shear layers in the wake of the valve housing. Specifically, the turbulent stress at the outside edge of the x^- lateral jet in the normal orientation was much higher and was more expansive than in any of the shear layers for the aligned orientation. This was likely a consequence of the lateral jet impingement on the concave wall. Conversely, when the valve was in the aligned orientation, there was more space between the jets and the concave wall, and the jets did not impinge on it.

Another significant difference in the turbulent stress was shown on the valve symmetry plane: in the normal orientation, high levels of turbulent stress developed between the central jet and the non-coronary sinus (on the y - z centre-plane). This did not occur for the aligned orientation, for which the turbulent stresses were very low on the valve symmetry plane (the x - z centre-plane). Figure 6.15 shows the velocity projections on the valve symmetry planes for the normal and aligned orientations at peak flow and $z/D = 0.40$. For the normal orientation, the “shoulder” of the central jet on the y^- side protruded, forming a local velocity peak in between the central jet “head” and the reverse flow in the non-coronary sinus. This resulted in a steep velocity gradient between this shoulder and the reverse flow, which increased

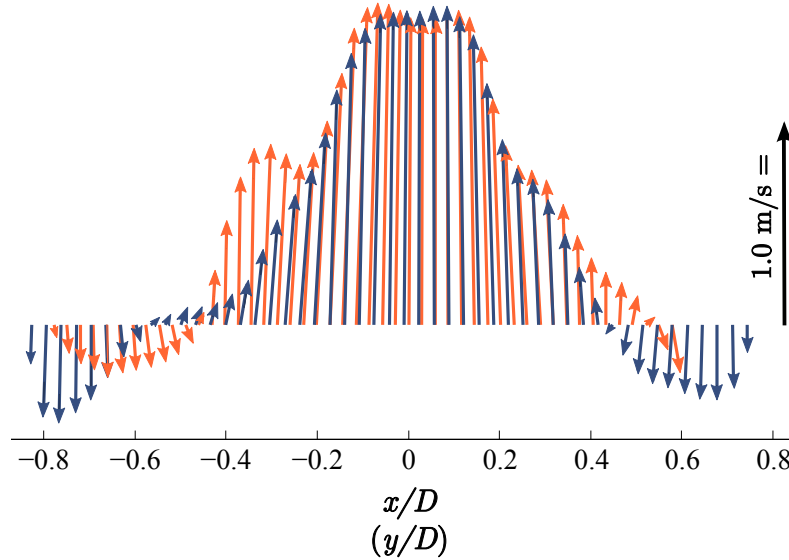


Figure 6.15: Velocity projections on the planes parallel to the valve symmetry plane for normal (orange arrows) and aligned (blue arrows) valve orientations at peak flow and $z/D = 0.40$.

the production of turbulence there. In the aligned orientation, on the other hand, the central jet had no prominent “shoulder” at all, and the velocity transitions between the reverse sinus flows and the central jet were much more gradual. The asymmetric protrusion of the central jet shoulder observed for the normal orientation may have resulted due to the asymmetry of the sinuses relative to the central jet on the valve symmetry plane. In contrast, for the aligned orientation, the channel walls of the left and right sinuses were relatively symmetric about the central jet profile.

6.4.3 Effects of valve orientation on right coronary flow

As was previously documented [86] and was also reported in Section 4.2, the lateral jets emerged before the central jet during flow acceleration. This was also apparent when comparing the coronary flows on the x - z plane for the different valve orientations (Figure 6.9). Consequently, the sinus vortex advanced earlier where it was adjacent to the lateral jet than where it was adjacent to the centre jet. It was found that, as a result, when a lateral valve orifice was oriented toward the coronary artery (*i.e.*, in the normal orientation) the sinus vortex blocked the coronary ostium earlier than for

the other valve orientations. This caused the coronary flow to be much lower than for the other orientations, during flow acceleration.

At mid-acceleration, the aligned orientation resulted in the highest coronary flow rate. In addition to the delayed advance of the sinus vortex, the coronary flow for this orientation was augmented by the high velocity fluid of the jet adjacent to it, part of which rolled over the sinus vortex into the artery. For the inclined orientation, the coronary flow also benefited from the delayed sinus vortex; however, the adjacent fluid emitted from the valve was not as fast and, consequently, the coronary flow rate was slightly lower than for the aligned orientation.

During flow deceleration, the coronary flow was highest for the inclined valve orientation. This appears to have been a direct result of the size, shape and orientation of the sinus vortex. For both the normal and aligned orientations, the sinus vortex next to the coronary ostium was confined by the adjacent lateral jet, which was aligned with the axial direction. For the inclined orientation, however, the fluid velocity in the aortic root had a significant transverse component directed away from the coronary ostium. As a result, the sinus vortex was confined by a lesser amount, and it incorporated more of the fluid from the channel centre. Additionally, this caused the long axis of the vortex to be inclined more than for the other orientations, which permitted the fluid from it that entered the coronary artery to be more in-line with the artery's axis. Consequently, there was a weaker flow separation at the artery entrance, and a larger portion of the artery contained forward moving fluid.

The measurements presented here reveal information about the three-dimensional shape and time evolution of the sinus vortex ring. The vortex had a larger cross-sectional diameter on the x - z plane at $y/D = 0.165$ where the right coronary sinus was deepest, compared to on the x - z centre-plane which was not as deep into the sinus. This was expected as it is presumed that the vortex would occupy the space between the valve housing and the channel wall, which is larger in the depths of each sinus than at the commissures between them. We can infer that the sinus ring would therefore be thicker in all three sinuses and thinner at the locations between them and that it would have a trimodal shape similar to the perimeter of the aortic root. This inferred shape is depicted in Figure 6.16. The cross-sectional diameter of the



Figure 6.16: Sketch of inferred vortex ring shape without the effects of valve orientation.

sinus ring was further affected by the valve geometry: it was larger where it was adjacent to the central orifice than where it was adjacent to the lateral orifices. The largest vortex ring diameter was thus achieved when the valve's line of symmetry was in-line with the right coronary ostium (the inclined orientation): in this orientation the portion of the vortex ring located within the right sinus was also adjacent to the valve's central orifice. As we saw in Section 6.3, this configuration resulted in the largest forward flow volume in the contiguous right coronary artery. Furthermore, the measurements illustrated that the sinus ring was advected downstream non-uniformly: it was advected earlier where it was adjacent to the lateral orifices than where it was adjacent to the central orifice. This would have caused the trimodal-shaped ring to bend in a "V-shape" about the valve's plane of symmetry, as was similarly reported by Dasi *et al.* [27].

It must be noted that the results from the measurements of coronary flow have limited relevance to physiological conditions. *In vivo*, the majority of coronary flow occurs during diastole, because the contraction of the heart during systole compresses the coronary arteries, increasing their flow resistance. During diastole, while the bulk flow rate in the aorta has decreased to nearly zero, the heart muscle relaxes and the coronary arteries dilate, causing blood to flow through them. This effect was not accounted for in the mock circulatory apparatus; to do so, an active, timed resistance would have had to be applied at the outlets of the coronary arteries, which would have

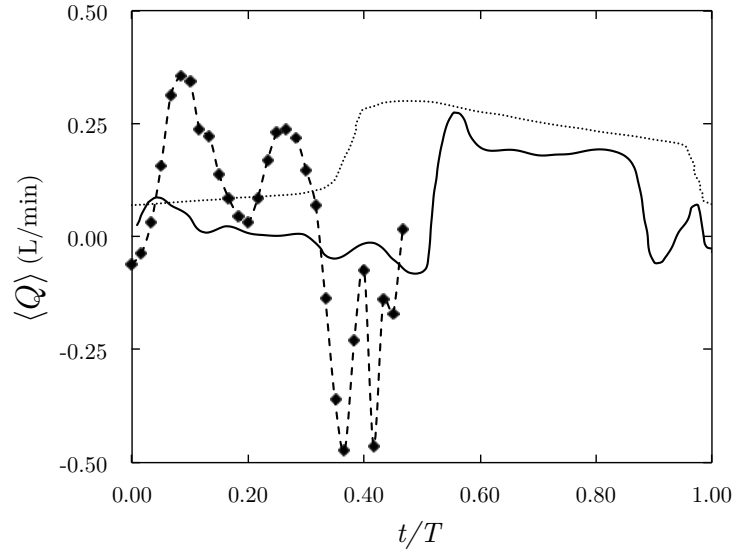


Figure 6.17: The flow rate in the right coronary artery for the normal valve orientation (diamonds) compared with physiological coronary flow measurements from de Tullio *et al.* [75] (solid line) and Hall [40] (dotted line).

been considerably difficult to implement. For this study, only a passive resistance was applied that was constant over the course of the cycle. As a result, the majority of the flow occurred during systole, rather than diastole. The coronary flow rate for the normal orientation is shown with measurements of physiological coronary flow from de Tullio *et al.* [75] and Hall [40] in Figure 6.17. Additionally, during diastole, the fluid in the coronary artery drained back into the aorta, creating a jet. This does not occur in healthy hearts, although it can occur in patients with severe aortic valve disease or coronary artery disease [17]. The present results reveal the complex interactions between the flow in the aortic root with those in the coronary arteries without the added effect of vascular resistance caused by intramyocardial stress. They provide a first step in understanding the physiological flow interactions and may be used in future studies, which account for the time-varying vascular resistance, as a basis for comparison in order to understand those effects.

In summary, the bulk flow rate in the right coronary artery during systole was highest when the valve was aligned so that its plane of symmetry intersected the artery opening. This was due to the fact that, during flow acceleration, the vortex in the

right coronary sinus was convected downstream later than for the normal orientation, and also due to the size and orientation of the sinus vortex during deceleration, which diverted more flow towards the artery from the channel centre and at an angle that was more axially aligned with it. While the flow rate was not measured in the left coronary artery for the three orientations, it is presumed that, although the inclined valve orientation was optimal for the right coronary flow, it would not be optimal for the flow in the left coronary artery. When the valve's symmetry plane is aligned with the right coronary ostium, one of the valve's lateral orifices is positioned next to the left coronary ostium; it was found that, for the right coronary artery, this arrangement resulted in the lowest coronary flow rate. One may speculate that the aligned orientation would be optimal for the flows in both coronary arteries, if these are considered together: it resulted in the next highest flow rate after the inclined orientation, and it resulted in the symmetric positioning of the ostia relative to the valve geometry, which means that the arterial flows should be equally affected. If a condition existed, however, where the flow in only one coronary artery was for some reason deficient, it could potentially be boosted by implanting the valve so that its symmetry plane were aligned with that artery.

Chapter 7

Conclusion

7.1 Summary of results

The flow characteristics past a bileaflet mechanical heart valve (BMHV) were investigated first within an axisymmetric channel and then within an anatomical model of the aorta. Velocity measurements were conducted within a mock circulation loop, which produced physiological, pulsatile flow conditions. Single-point and planar velocity measurements were made, and the turbulent and viscous stresses were calculated as metrics of potential blood damage.

In both channels, a triple jet developed past the valve, and reverse flow occurred in the sinus region between the jets and the channel wall. The measurements made in the axisymmetric channel matched very closely those made previously in similar channels. The single-point measurements were in good agreement with the planar velocity measurements.

The velocity fluctuations, and both the turbulent and viscous stresses were highest at the shear layers that developed off the leaflet tips and the valve housing. These regions corresponded to the edges of the three jets that were emitted from the valve. Maximum and average stress levels were higher on the planes perpendicular to the valve symmetry plane than on those parallel to it. The viscous stresses were highest closest to the valve and decreased downstream. On planes both parallel and perpendicular to the valve's symmetry plane, the viscous stresses were higher on off-centre planes that were located at the jet edges than on the centre-planes. On the planes perpendicular to the valve's symmetry plane, the turbulent stresses increased downstream of the valve, and were maximum near the end of the expanded "pseudo-sinus".

Measurements were made past the BMHV within the anatomical aorta model, first with the valve positioned so that its plane of symmetry was perpendicular to the

plane of aorta curvature. This was called the “normal” orientation. The outlets of the branching arteries were initially clamped so that there was no flow through them. These measurements were compared to the ones made in the axisymmetric channel to determine the effects of the anatomical aorta geometry on the flow characteristics past the valve.

In general, the flow characteristics closely downstream of the valve and near the valve axis were very similar for the two channel geometries. Large differences, however, emerged further downstream and at radial distances away from the channel axis. The most significant differences occurred at the outside edges of the lateral jets, where both the turbulent stresses and the viscous shear stresses were found to be much lower in the anatomical than the axisymmetric model. This was attributed to geometrical differences of the sinuses: the anatomical sinus had a lower height-to-width ratio which caused less stretching of the sinus vortex, and it had inclined walls which allowed a more gradual reorientation of the flow.

The velocity distribution past the valve was affected by the curvature of the aorta: the velocity was skewed toward the convex wall during flow acceleration, but toward the concave wall during flow deceleration. The aorta curvature was not found to cause increased fluid stresses compared to the straight channel, and the turbulent and viscous stresses were lower in the anatomical model.

Measurements were made for two additional valve orientations: with the valve’s plane of symmetry (1) aligned with the aorta’s plane of curvature, called the “aligned orientation”, and (2) aligned with the opening to the right coronary artery, called the “inclined orientation”. Measurements were first made with no flow through the branching arteries and then with flow through the right coronary artery for all three valve orientations.

When the valve was in the normal orientation, a lateral jet impinged on the concave wall just downstream of the end of the sinus, due to its position relative to the aorta curvature. Consequently, the turbulent stresses associated with this jet were higher, but more confined than for the other lateral jet, for which the associated stresses extended far into the ascending aorta. It was found that the turbulent stresses in the sinus region past the valve were 20% higher when the valve was in the normal

orientation compared to the aligned one. Additionally, the occurrence of high-level viscous stresses (≥ 10 Pa) was 30% higher for the normal orientation than the aligned one. This was attributed to (1) the impingement of a lateral jet on the concave wall of the ascending aorta, and to (2) steeper velocity gradients measured parallel to the valve symmetry, which both occurred when the valve was “normally” oriented to but not when it was “aligned” with the aorta’s symmetry plane.

For the normal valve orientation, a low velocity region extended from the non-coronary sinus into the ascending aorta and persisted throughout peak flow. This could have clinical consequences, as such regions can foster thrombus growth and, by altering the normal shear stress distribution at the adjacent vessel walls, promote arterial disease. Positioning the BMHV as far as possible into the non-coronary sinus at implantation may minimize the size of this low velocity region. For the other two valve orientations, it was not possible to determine the full extent of this region; however, the results indicated that this low velocity region would be less extensive than for the normal valve orientation.

From the measurements for the inclined valve orientation, high levels of turbulent stresses were found to issue from the corners of the valve’s lateral orifices, where shear layers from the valve housing and the valve leaflets occurred. These levels were higher than at any location measured for the other two valve orientations, for which the measurement planes were orthogonal to the valve’s symmetry plane. The viscous stresses were also relatively high past the orifice corners, but they were less than the maximum viscous stresses measured on the planes perpendicular to the valve symmetry plane, for the other orientations.

Measurements were made of flow through the right coronary artery for the three valve orientations, and the bulk flow rate in the coronary was calculated for the forward flow phase. For all valve orientations, the majority of coronary flow occurred at the mid-acceleration and the mid-deceleration phases of the flow cycle. The coronary flow substantially decreased at peak flow, due to the blockage of the coronary ostium by the sinus vortex. The total flow through the artery was highest for the inclined valve orientation. For the normal orientation, the sinus vortex was advected downstream and blocked the coronary ostium earlier than for the other orientations, which

resulted in the lowest coronary flow volume. For the inclined orientation, the sinus vortex extended further into the channel centre, drawing more flow into the coronary. Additionally, its axis was more inclined than for the other orientations which aligned the fluid entering the artery more with the coronary axis, resulting in a smaller region of flow separation.

It should be noted that the results of this study were obtained for one patient's anatomy, and that there may be some variation in the flow characteristics from one patient to the next due to the variation in their specific anatomies. It was for this reason that the results were interpreted in relation to general anatomical features that most all people have, *e.g.*, three sinuses at the aorta's base, coronary arteries that branch from the sinuses, and curvature in the ascending aorta and arch. I expect that, although minor differences in the flow patterns would inevitably exist between patients, the general findings would be applicable for all patients; *e.g.*, positioning the valve normal to the aorta's curvature would likely result in jet impingement and the associated increase in fluid stresses, and aligning the central orifice of the valve with a coronary ostium would result in the sinus vortex being larger next to the ostium and increase the volume of fluid entering it. Even so, the results of the study are limited by not being able to match all physiological parameters, specifically, the anatomical compliance and the time-varying resistance of the coronary arteries. Further studies are required to determine these effects.

Table 7.1 summarizes the results for the axisymmetric model and for the three valve orientations in the anatomical model. Values of average and peak turbulent stresses RS_{max} and the percentages of viscous stress measurements exceeding 10 Pa are presented for the x - z and y - z centre planes, and for the combined measurements on both planes. The forward flow volume measured in the right coronary artery is presented for the three valve orientations.

7.2 Summary of contributions

To our knowledge, this is the first experimental study to measure either the turbulent or viscous stresses past a BMHV within an anatomical model of the aorta, and the

	plane	axisymmetric model	aorta model		
			normal	aligned	inclined
RS_{max} (Pa)	x - z centre	33 (172)	18 (91)	14 (64)	(205)
	y - z centre	22 (90)	19 (76)	17 (84)	NA
	combined x - z , y - z	28	19	15	NA
τ_{max} (% ≥ 10 Pa)	x - z centre	4.0	1.7	1.3	NA
	y - z centre	0.3	1.8	2.7	NA
	combined x - z , y - z	2.2	1.7	1.3	NA
right coronary flow volume (mL)	x - z , $y/D = 0.165$	NA	0.74	0.84	1.2

Table 7.1: Summary of results: values, where applicable, of turbulent stress RS_{max} , percentage of measurements for which $\tau_{max} \geq 10$ Pa, and systolic right coronary flow volume for measurements made in the axisymmetric model and for the three valve orientations in the anatomical model. The plane-averaged values of RS_{max} are shown as nominal, with the plane-maximum values shown in parentheses.

first study of any kind to measure both of these stresses in a model that included the arteries that branch from the aorta. Measurements were made under controlled and repeatable flow conditions, which closely approximated physiological flow. High quality measurements were made on numerous planes and in sufficient quantities to achieve statistical convergence.

Prior to acquiring measurements in the anatomical aorta model, measurements were prudently acquired in an axisymmetric model to provide a basis for comparison. Because the measurements were made within the same apparatus and under nearly identical flow conditions, this permitted us to quantify the effects of specific anatomical features on the flow characteristics past the valve. Of particular interest, it was found that the geometry of the anatomical aorta sinus was effective in reducing turbulence compared to the straight, axisymmetric geometry which is commonly used to model BMHV flows. An important application of this finding is toward aortic root replacement. In patients for whom the structural integrity of their aorta has deteriorated resulting from an aneurysm or from a disorder of the connective tissue called Marfan's syndrome, the aortic root may need to be entirely replaced with a

graft. The aortic valve may sometimes be preserved, or it may be replaced with a prosthetic valve [29]. The graft may consist of a straight tube, or it may have an axisymmetric expansion to act as a “pseudo-sinus” [30]. The present finding suggests that, in order to reduce the turbulence within it, the shape of the “pseudo-sinus” should be as similar as possible to that of the natural sinus; specifically, it should have an height-to-width ratio close to 1, and it should have inclined walls.

By quantifying the effects of the anatomical aorta geometry on BMHV flow characteristics, the results of this study may be used towards determining for future studies which geometrical simplifications of the aorta may be acceptable to adopt. Moreover, they will provide an understanding of the limitations to each simplification. For example, it was found that the fluid stresses were more significantly affected by the sinus shape than by the aorta curvature; considering this, an investigator who is only interested in the fluid stress levels might choose to use a channel with an anatomically realistic sinus, but with a straight tube downstream of it.

The configuration used for this research enabled us to measure the effect of valve orientation on the downstream flow characteristics; it was found that lower turbulent and viscous stresses resulted when the valve was aligned with the plane of aorta curvature. Only two other studies are known to have investigated the effect of BMHV orientation on fluid stresses: an *in vivo* study by Kleine *et al.* [47] and an *in silico* study by Borazjani & Sotiropoulos [13]. Kleine *et al.* [47], using Doppler ultrasound, obtained measurements at a few discreet locations past the valve, thus missing large regions of the flow, whereas the present study encompassed a much larger portion of the flow domain. The numerical study by Borazjani & Sotiropoulos [13] calculated the flow of the 3D flow domain past the valve, but their model did not include any of the branching arteries, and they did not calculate the turbulent stresses.

It has long been known that the vortices that form in the aortic sinus are implicated with the flow in the coronary arteries, but the specific interactions between them have been little understood. This was the first study to measure the coronary flow rate for varying BMHV orientations and to also characterize the interactions between the flow structures in the aortic root and the coronary artery. The measurements revealed how, in the absence of intra-myocardial pressures acting on the

coronaries, the coronary flow is highly dependent on the size, shape, orientation and spatial location of the sinus vortex. It was also found that these characteristics of the sinus vortex and, consequently, the coronary flow rate were sensitive to the orientation of the bileaflet valve.

The findings of this research can assist surgeons in choosing the best implantation orientation for a BMHV. In order to have lower stresses on blood, it is recommended that the valve be implanted with its plane of symmetry aligned with the aorta curvature. This same orientation is also recommended for maximum flow rates in both coronary arteries. On the other hand, if a patient has one coronary artery that is flow deficient, then the flow in that artery could be potentially increased by aligning the valve with its ostium.

7.3 Recommendations for future work

The planar PIV system used for this research measured only two components of the velocity vector in a three-dimensional flow. This resulted in underestimation of the turbulent and viscous stresses. Additionally, measurements were limited to a few planes and did not include planes parallel to the valve. Future measurements in these transverse planes could provide information about the secondary flow structures that develop as a consequence of the aorta curvature and their influence on BMHV flow characteristics. Future studies may also use stereoscopic PIV to obtain the third velocity component and provide better estimates of the stresses.

In order to precisely control the flow conditions at the inlet of the BMHV, a straight channel geometry was used upstream of it. In the future, a more realistic geometry of the ventricle could be used upstream of the BMHV in order to generate more physiological inlet conditions and to study ventricular effects; however, precise control of its contraction should also be replicated.

For the measurements of flow in the coronary arteries, a constant resistance was applied to the artery outlets. In the heart, the coronary arteries are subjected to time-varying intra-myocardial pressure, which changes their flow resistance over the course of the cycle. In future work, such a time-varying resistance could be applied

to the artery outlets to better simulate *in vivo* conditions. Additionally, the present measurements made with PIV were acquired only at phases during systole. Future studies of coronary flow with time varying resistances should also measure the flow during diastole, as the majority of coronary flow normally occurs then. Such a study could use the present results to quantify the effects of intra-myocardial pressures on coronary flow patterns and flow rates.

The mock circulation loop designed and built for this research generates physiologically accurate pressure and flow waveforms. It is fully adjustable to replicate a large range of healthy and diseased flow conditions and is suitable for use with a variety of flow visualization and velocity measurement techniques. It could be used for a multitude of future studies of cardiovascular flows; for example, different types of valves may be tested within it, including bioprosthetic valves and those used for transcatheter aortic valve implantations (TAVI). Flows in different blood vessels may also be studied, such as, for example, the blood vessels of the brain.

Appendices

Appendix A

Flow Setting Protocol, Laser Alignment and Calibration Procedures

A.1 Mock circulation loop set up

The following procedure was used to adjust the pressure and flow waveforms in the test section produced by the mock circulation loop (MCL). It was performed each day, prior to acquiring LDV and PIV measurements. The MCL settings, and the pressure and inlet velocity waveforms were carefully monitored at regular intervals throughout measurement acquisition. The following procedure assumes that the mock circulation loop (MCL) has been filled with the blood analog fluid, with the proper test section and valve installed. Various components of the MCL are illustrated and labelled in Figure A.1.

- Turn on the MCL power bar, connect the data acquisition device (DAQ) to the computer, and open the compressed air ball valve.
- Ensure that the fluid reservoir is open to the atmosphere (its ports may be closed in between experiments in order to prevent fluid evaporation).
- Check that the compressed air pressure in the main air line is 24 psi. Adjust this using the pressure regulator at the wall if necessary.
- Check the air volume in the compliance chamber C_1 : the fluid level should be at the marked line that is 3.1 cm from the top of the chamber. If the level is above this line, air must be reintroduced to the chamber by operating the loop with the valve R_2 closed and the downstream compliance chamber open to the atmosphere. Run the loop until the fluid reservoir empties and air is pumped

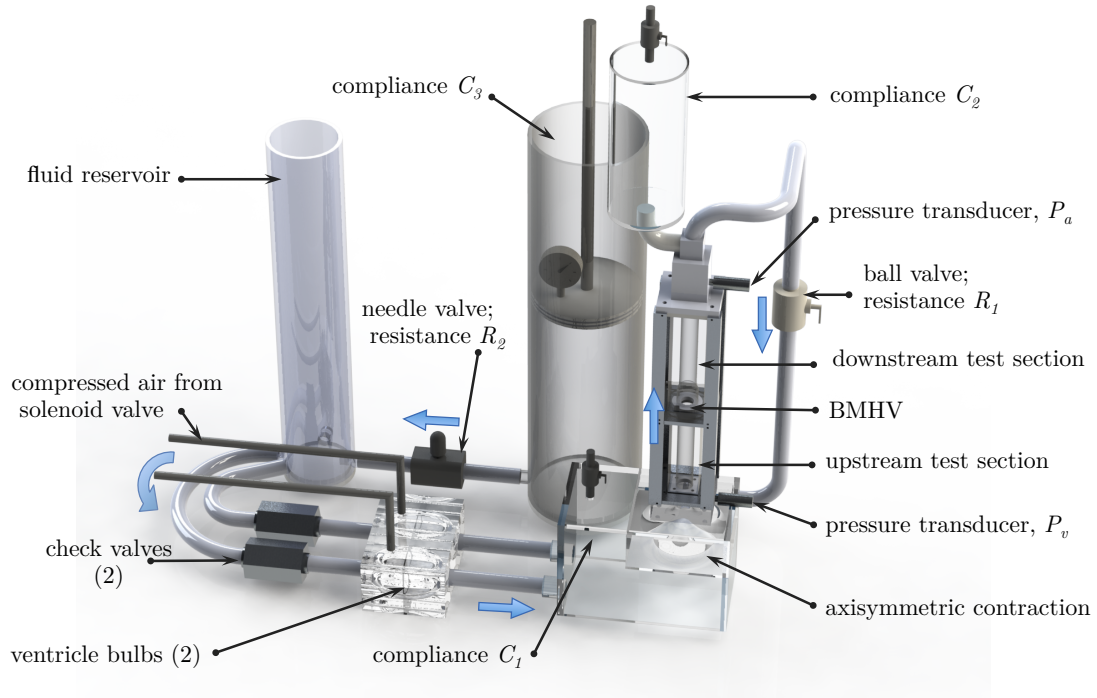


Figure A.1: Components of the mock circulation loop.

into the compliance chamber C_1 . To raise the fluid level in this chamber, air can be let out through the venting valve.

- Check the position of the valve R_1 : the valve handle should be positioned at 52 degrees. (this position is rarely changed, but slight changes in its position can have significant effects on the flow).
- Check the position of the valve R_2 : the needle valve should be set to “9” on the “brown-grey” color band rotation.
- Measure the fluid level in the compliance chamber C_3 : it should be 35 cm from the bottom of the chamber, but this level decreases if fluid is lost or evaporates over time. If fluid loss is significant, additional fluid should be added to the loop.
- Record the pressure P_3 in the compliance chamber C_3 : the level should read 0.35 psi on the gauge in the compliance chamber. If the pressure is lower than

this, add air to the downstream compliance chamber with a bicycle pump until the correct level of P_3 is obtained.

- Open the LabVIEW program to read the pressure downstream of the BMHV, P_a . Run the loop for several minutes until steady state is reached. Steady state is determined when there is no visible change in the pressure curve from one pulse to the next.
- Record the dynamic value of the pressure to the ventricle bulbs P_{bulb} . Carefully watch the moving needle during several pulse cycles to record the highest value over the cycle; it should be approximately 7.2 psi. Adjust the pressure using the “bulb pressure regulator” as required.
- Check the fluid level in the compliance chamber C_2 . While the loop is not operating, this level is below the bottom of the chamber, but when the loop starts to run, the system becomes pressurized causing the fluid level to rise into the chamber. At steady state, the level should fluctuate about a line that is marked with a pen on the chamber wall (17 cm from the top of the chamber).
- In LabVIEW, visually check that the pressure curve is in the correct range, and has the correct shape. Record the pressure for 10 to 15 pulses.
- In Excel, open the data file of the recorded pressure. Identify the start and end times of a single pulse from the change in the output voltage of the solenoid valve. In a new column in Excel, calculate the reference time t' of each measurement by subtracting the start time of the cycle from the absolute measurement times. Plot P_a vs. t' , together with a typical pressure curve obtained on a previous measurement day (it is best to use the same reference pressure waveform each time).
- With the LDV laser aligned (following the alignment procedure in Appendix A.2), position the laser volume to a location where a previous LDV measurement was taken (usually a centre-line location $[0,0,z]$, where z is the axial location of the last profile measured).

- With the LDV laser turned on and warmed up, the BSA software and a project file opened, and the MCL running at steady state, adjust the BSA parameters to achieve a good data rate and data validation as indicated by the “system monitor”. Hit “run continuously” to see the plots of the measured velocity and histograms. Once a good data rate and validation are achieved (several hundred Hz, and $> 90\%$, respectively), record the velocity for approximately 200 cycles. Export the recorded data to Excel. Plot the ensemble average of the axial velocity $\langle W \rangle$ *vs.* the cycle time. Compare the velocity waveform with a reference waveform from data taken on a previous day at the same location.
- If the pressure or flow waveforms differ slightly from the reference waveforms, verify the loop parameters to see what might need adjustment. This is most often: the pressure to the ventricle bulbs P_{bulb} , due to pressure changes elsewhere in the compressed air line; pressure P_3 , due to air leakage out of the C_3 compliance chamber over time; or the air volume in the C_3 compliance chamber, due to air leakage. If the MCL has been filled with the blood analog fluid for an extended period of time ($\geq$ weeks), air leakage out of the C_1 chamber may be great enough to affect the flow and air may need to be reintroduced to the chamber by the method described previously. The positions of the resistance valves R_1 and R_2 should not have changed from previous measurements.

A.2 Laser alignment for LDV

The following procedure was performed to align the LDV laser to the test section. It was performed at regular intervals throughout each measurement series; for example, at every change in z location.

- Turn on LDV laser and wait for it to warm up (20-30 minutes).
- Tape a strip of laminated grid paper to the front panel of the test section. The strip should sit flat against the top of the BMHV housing flange. Align the vertical edge of the grid paper with the vertical edge of the test section. The

grid paper should indicate millimetres. This placement is temporary, and only for laser alignment.

- Roll alignment: Check that pairs of vertical and horizontal laser beams align with the vertical and horizontal directions on grid paper. If they do not, loosen the screws on laser mount and rotate probe body until the beams align with the vertical and horizontal directions. Re-tighten the screws on the laser mount.
- Pitch alignment: Using a level, check that the laser probe body is mounted horizontally. If it is not, loosen the bolt that attaches the mount to the traverse, and adjust using metal shims, then re-tighten the bolt. Check that the plane that the valve is seated in is also level (i.e. the inlet flange of the downstream test section).
- z -direction alignment: Move the traverse up or down in order to position the horizontal beams to the desired z axial position marked on the grid paper.
- y -direction alignment: Move the traverse in the x -direction so that the intersection volume is approximately centred. With the laser powered on low (50 mW), move the traverse in the y -direction so that the laser intersection volume intersects with the nearest, and then the farthest wall. Identify when the intersection point is at the wall by looking through a microscope that is mounted on the traverse and focused on the intersection volume. Record the y -dial readings at each wall. Zero the y -dial to equal the mid-point, and repeat until no further adjustment is required.
- x -direction alignment: Set up a dial gauge (with a 3" range) so that it is clamped to the rails of the traverse, with the indicator depressed approximately half of its range and in contact with a flat surface of the moving laser mount. Align the gauge so that the indicator is horizontal to the ground and runs parallel with the test section. Move the traverse along the rails in the x -direction, so that the laser intersection volume intersects with the right, then the left tube wall (when the measurement volume is located at the channel wall, the the laser

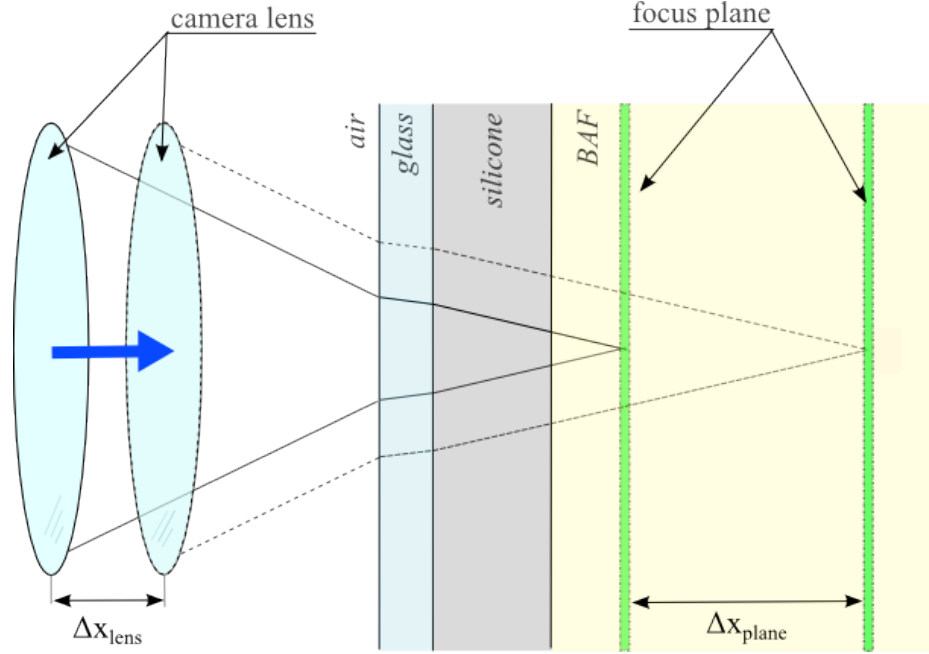


Figure A.2: Sketch of the refraction of light rays as they pass through the divers media of the test section. The phenomena results in a difference between the camera lens displacement Δx_{lens} and the corresponding displacement of the focus plane Δx_{plane} . (Sketch is not to scale, and light ray angles are exaggerated for demonstration.)

beams past their intersection point “disappear”). Note the reading on the dial gauge and repeat for the other wall. Set the zero of the dial gauge to equal the calculated mid-point. Repeat until no further adjustment is required.

A.3 PIV calibration

Due to the diffraction of light as it passed through the test section, the displacement of the camera lens in the direction normal to the test section wall was not equal to the corresponding displacement of the focus plane. This effect is depicted in Figure A.2. Using Snell’s law of diffraction with the refractive index of the test section and the working fluid, it was calculated that for every millimetre the camera moved towards or away from the test section, the focus plane moved 1.41 mm in the same direction. This ratio of optical displacement to physical displacement will be designated α_η .

The system calibration procedure accounted for this difference between camera and focal plane displacement. The steps are as follows:

1. A sheet printed with dots arranged in a grid (0.8 mm in diameter, and spaced 2 mm apart) was aligned with, and fixed to the outer wall of the test section that was between the measurement plane and the camera (the calibration wall).
2. The position of the desired measurement plane was carefully measured and marked on the test section walls, at three points that fully defined the plane. The distance between the measurement plane and the calibration wall is designated d_{pl} .
3. With the laser head and camera mounted and levelled on the traverse, the traverse and mirror were positioned to visually align the laser sheet so that it hit the marks made in step 2.
4. The traverse was moved the distance d_{pl}/α_η , in the direction that moved laser sheet towards the calibration wall.
5. The camera was manually focused on the dots on the calibration wall, an image was taken, and the the calibration was performed using an algorithm from the LaVision software.
6. The traverse was moved back the distance d_{pl}/α_η , to where laser sheet hit the target marks of the measurement plane. Here, the particles were in good focus, but the focus of the camera was fine-tuned so that the particle images were sharpest, with minimal diameters.

Appendix B

Selection of the Bileaflet Mechanical Heart Valve

During valve replacement surgery, the bileaflet valve is sized based on the diameter of the patient’s aortic annulus. The annulus is measured using a barrel sizer, which should pass easily through the annulus with no resistance. The largest size that can easily pass is selected [78].

The aorta model for use in this study does not subsume an annulus. The model was made using an aorta from a cadaver which, excised from the body, collapsed open where it would normally be cinched in at the annulus. In the absence of an annulus, other geometrical features of the aorta were used to determine a suitable annulus size, and ultimately, the size of the valve.

Various studies have used two-dimensional echocardiography to document normal aortic root dimensions in children and adults, [56, 60, 69, 90]. The most common standard of measurement is the method proposed by Roman *et al.* [60]. Using this method, two-dimensional measurements are taken at various features of the root in parasternal long-axis view. Roman *et al.* took measurements at four levels: the annulus, the sinuses of Valsalva (at the largest diameter), the sinotubular junction, and the proximal ascending aorta. The locations of these measurements are illustrated in Figure B.1. Measurements were made using the “leading edge technique” from views showing the largest diameters (Figure B.1).

It was decided that the sinotubular junction (STJ) diameter would be the feature most representative of the size of the aorta model and was used to determine an appropriate annulus size. A male mold of the aorta root was made by pressing modelling clay into its entrance. A square piece of latex was used as a barrier between the clay and the model to protect the model from clay residue. The male mold was baked to harden, and then measured carefully at its sino-tubular junction as well as at the sinus of Valsalva. Measurements were taken across what would be the

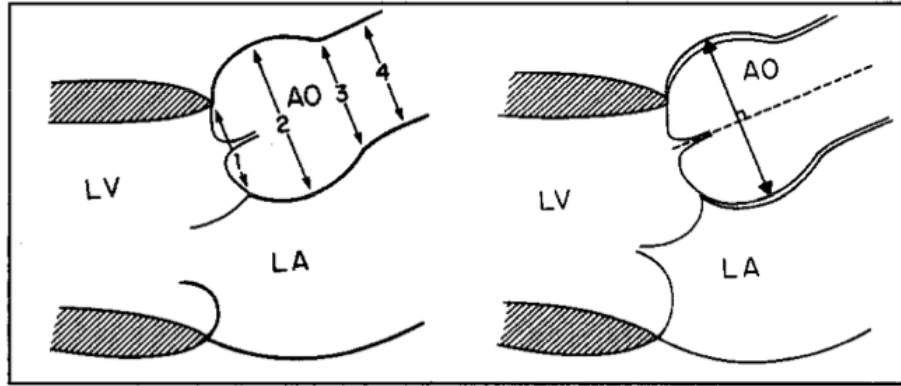


Figure B.1: Left: levels of measurement of the aortic root by Roman *et al.* 1) the annulus; (2) the sinuses of Valsalva; (3) the sinotubular junction; and (4) the proximal ascending aorta. Right: measurements made using the leading edge technique. Ao = aorta; LA= left atrium; LV = left ventricle. (Reproduced from [60])

parasternal long-axis view (from the right coronary sinus across to the non-coronary sinus). Averaging five measurements each, the STJ was determined to be 24.3 mm, and the sinus diameter to be 33.9 mm. The average ratio of the STJ to the aortic annulus was reported by Roman *et al.* [60] to be 1.1 ± 0.1 , and by Zhu *et al.* [90] to be 1.18. From the average of these ratios, an appropriate diameter of the aortic annulus for our model is calculated to be 21.3 mm. It was thus concluded that a 21 mm SJM RegentTM valve was suitable for our study.

It is interesting to note the relative size of the sinuses of the aorta model compared to the sinotubular junction: the ratio of STJ to sinus diameter being 0.72. A study by Tamas *et al.* [69] reports this ratio to be 0.75. Based on this, the sinuses of our model were 5% larger than average for its STJ size, which may be considered to be within the normal limits due to physiological variability.

Appendix C

Sinus Remodelling: Procedure for the Fabrication of the Modifying Insert.

The following steps were performed to fabricate the modifying insert. The steps and intermediate parts are illustrated in Figure C.1.

1. A male part of the existing aortic root, called part $1m$, was made using modelling clay (as was described in Appendix B).
2. A female mold, part $1f$, was made of part $1m$ using latex rubber.
3. A plaster support for part $1f$ was formed.
4. $1f$ was filled to the location of the maximum sinus diameter (6 mm from the top) with epoxy, forming part $2m$.
5. The bottom of the sinuses were fashioned out of modelling clay (part $3m$) to mate with $2m$. This was done by tracing the bottom surface of part $2m$ onto a block of clay approximately 10 mm thick and cutting out the profile. Then, the edges of one end were pressed inwards to form the annulus, approximately circular in cross-section with a diameter of 21 mm.
6. Part $3m$ was mated to $2m$, and they were placed inside the female mold, $1f$.
7. A “casting box” was built around the mold inlet from an acrylic tube with an internal diameter of 2.5”.
8. Liquid silicone was poured over the assembly, filling the void between parts $3m$ and $1f$, forming the sinus modifying insert. The mold was filled to a height of 4 mm above its inlet.

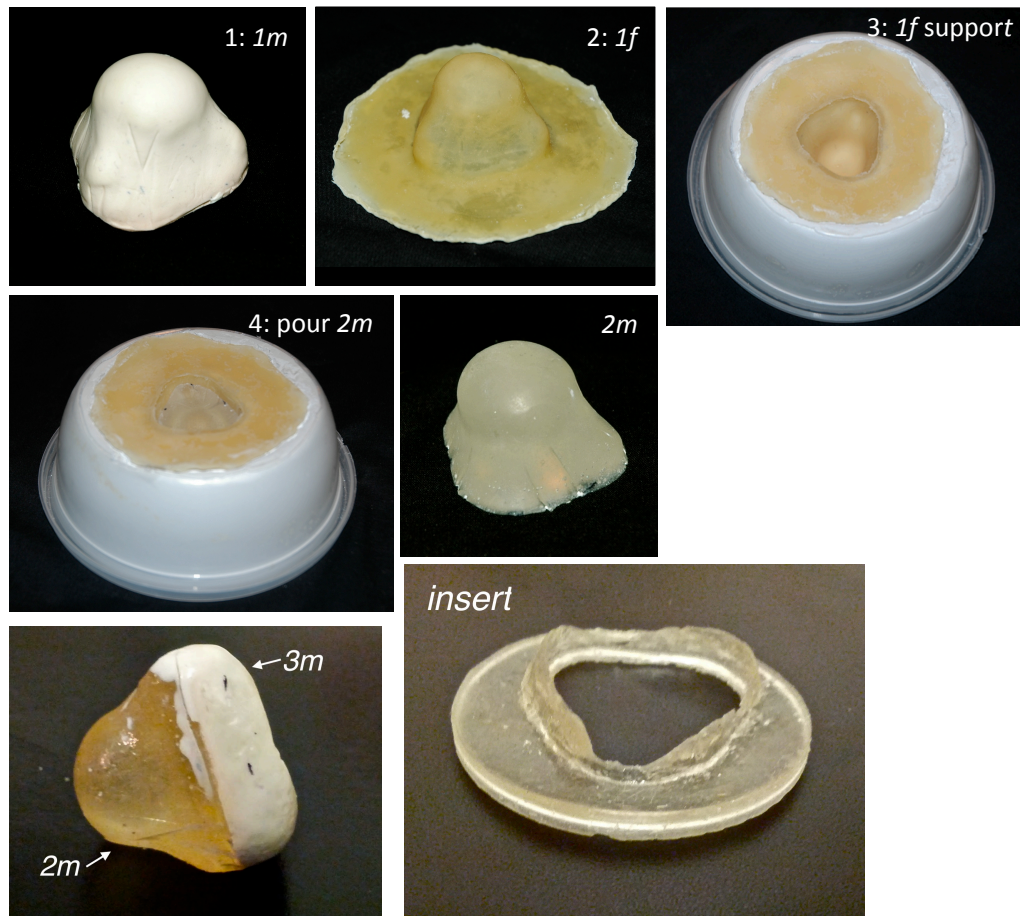


Figure C.1: Steps performed and intermediate parts made for the fabrication of the sinus modifying insert.

9. After curing, the insert was removed from the mold, and any rough edges were trimmed with a blade.

Appendix D

Procedure for Removing Outliers From Single-Point Measurements

The single-point measurements made with LDV were scanned to identify outlier measurements, which were discarded before further data processing. A measurement was identified as a potential outlier if it differed substantially from the mean of its bin. The criterion used to detect if a value U_{ik} was an outlier was

$$t_{ik} = \frac{|U_{ik} - \langle U \rangle_k|}{u'_k} \geq t_{cut-off} \quad (\text{D.1})$$

where cut-off value $t_{cut-off}$ was selected as the maximum number of standard deviations by which a value could differ from the mean without being considered an outlier.

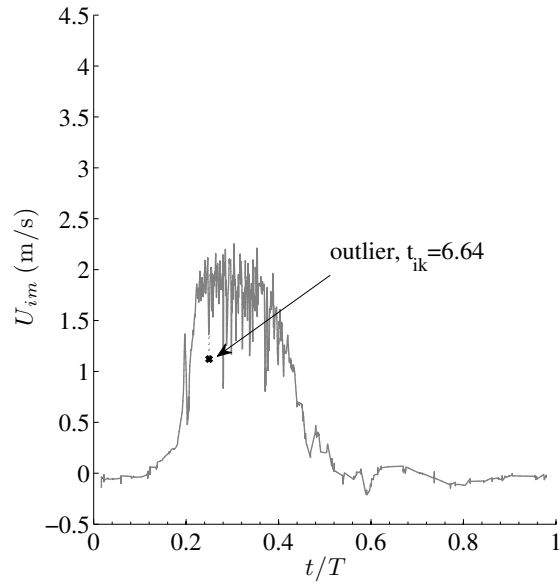
Chauvenet's criterion states that "if the expected number of measurements at least as deviant as the suspect measurement is less than one-half, then the suspect measurement should be rejected" [71]. Within each bin with N_{bk} measured values, the number of values that are expected to be as deviant from the mean as a measurement $U_{ik} = \langle U \rangle_k \pm t_{ik} u_k'$ is equal to N_{bk} times the probability that either $U_{ik} \leq \langle U \rangle_k - t_{ik} u_k'$ or $U_{ik} \geq \langle U \rangle_k + t_{ik} u_k'$ for any U_{ik} . This probability can be easily computed, if one assumes that the measurements in the bin have a Gaussian distribution.

The use of Chauvenet's criterion is not universally accepted among researchers, and some believe that data should never be rejected without firm evidence that they are false [71]. For the LDV measurements of this study, the criterion is used in a conservative manner, to reduce the probability of removing valid measurements. Following a manual inspection of the outliers that were identified by the use of different cut-off values, $t_{cut-off}$ was chosen to be 5.5; measurements having t_{ik} greater than this value were visibly obvious outliers, whereas measurements with t_{ik} less than this cut-off appeared to be potentially valid measurements.

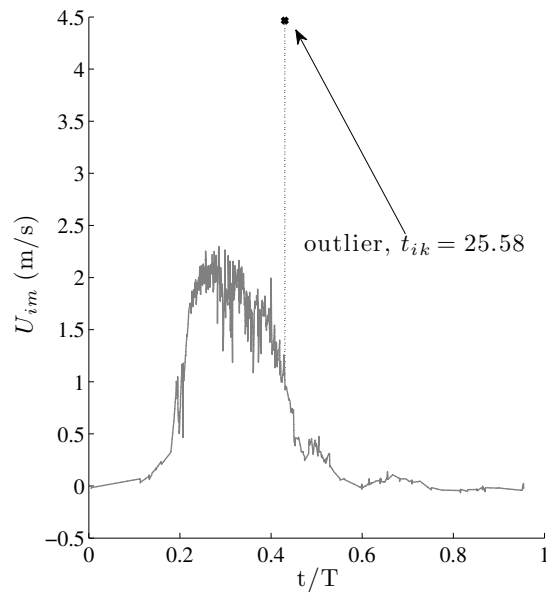
Assuming that the measurements are normally distributed, the probability that a measurement would fall outside of 5.5 standard deviations from the mean is $3.8\text{E-}8$. For $N_{bk} = 18,000$, which is a typical maximum number of measurements contributing to a bin, the number of measurements likely to exceed this cut-off is 0.0007, which is much less than Chauvenet's requirement of 0.5. This choice essentially prevents the elimination of valid data, while also detecting obvious outliers.

The identification and removal of outliers was achieved by using an in-house program written in MATLAB. Of the 1101428 measured values of U collected at location $[0,0,8]$ in the axisymmetric tube, 393 measurements (0.4%) were identified as outliers. Figure D.1 illustrates two of these cycles that were found to contain outliers. The outlier measurements were identified by the MATLAB program as having associated t_{ik} values of 6.6 (a) and 25.6 (b), which exceed the cut-off value of 5.5.

The effect of the outliers on the r.m.s. streamwise velocity fluctuations u'_k is illustrated in Figure D.2, which compares u'_k for position $[0,0,8]$ calculated before and after the removal of the outliers. The percent difference between the two values is near zero for most of the cycle, except at several discrete phases of the cycle where the percent difference has spikes. These spikes correspond with spikes of u'_k in the original data, which disappear when the outliers are removed. This clearly illustrates that there is a strong effect of outliers on the measurement statistics, and justifies their removal from the data set.



(a)



(b)

Figure D.1: Two representative cycles of LDV U measurements at location $[0,0,8]$, identified as having outliers; (a) cycle 12 and (b) cycle 249.

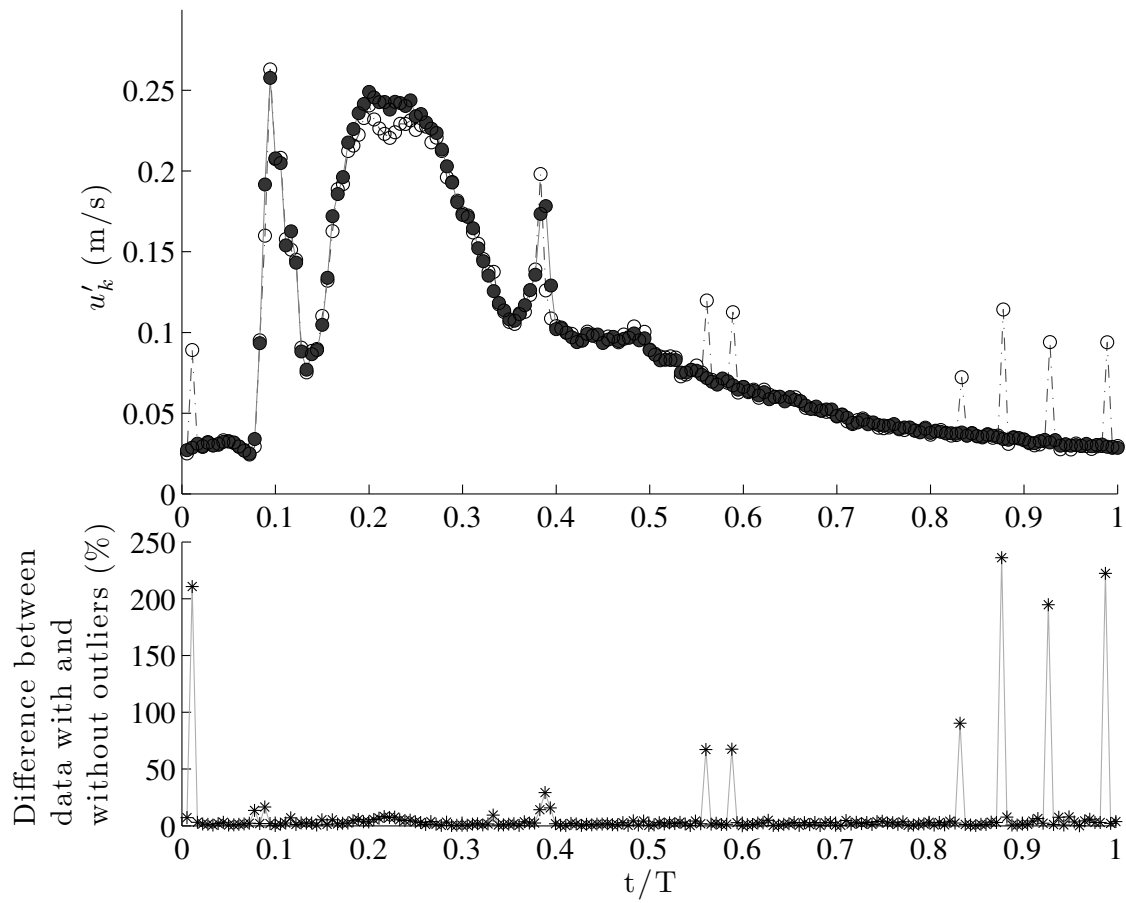


Figure D.2: Comparison of u'_k at position $[0,0,8]$, calculated with and without outliers. Results with $\circ$ original measurements, $\bullet$ outliers removed, and $*$ the percent difference between the two data sets.

Appendix E

Details of the Viscous Shear Stress Calculation from Planar Velocity Measurements

E.1 Grid refinement test

Figure E.1 illustrates the effects of final interrogation window size and the Gaussian filter on the calculated velocity. An axial velocity profile from the axisymmetric model was plotted for one instant during phase B of the cycle (mid-acceleration), having been calculated in four different ways: (1) using an 8×8 pixel final interrogation window size, (2) as in (1) but with a Gaussian filter applied, (3) using a 12×12 pixel final interrogation window size, and (4) as in (3) but with a Gaussian filter applied. The flow at this mid-acceleration phase has been previously described as laminar and repeatable [27, 74], so the velocity profile should not exhibit turbulent fluctuations, and any fluctuation observed in the profile could be considered to be noise. The level of noise for both of the 8×8 window size cases and for the 12×12 unfiltered case was unacceptably high and would have introduced artificially high velocity gradients that would have resulted in a significant overestimation of the viscous shear stress, but the 12×12 window size with the Gaussian filter resulted in acceptably low levels of noise; therefore these conditions were used for the calculation of the axisymmetric vector fields.

E.2 Corrective mapping algorithm

In Figure E.2, the phase-averaged τ_{max} resulting from two vector resolutions are compared: (1) from a 12×12 pixel interrogation window, denoted as τ_{max12} (Figure E.2 (a)); and (2) from a 24×24 pixel interrogation window, denoted as τ_{max24} (Figure E.2 (b)). It is clear that τ_{max24} is much lower than τ_{max12} . It is assumed that, because the

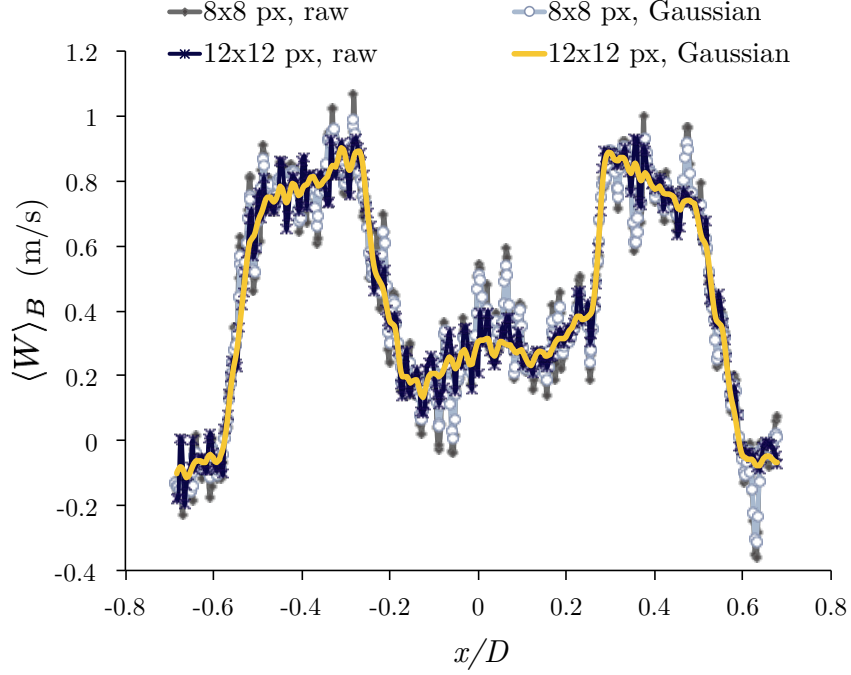


Figure E.1: Instantaneous axial velocity W at $z/D = 0.35$, during phase B, in the axisymmetric model, calculated in four different ways: (1) using an 8×8 pixel final interrogation window size, (2) as in (1) but with a Gaussian filter applied, (3) using a 12×12 pixel final interrogation window size, and (4) as in (3) but with a Gaussian filter applied.

noise was low for the vectors calculated with the 12×12 pixel interrogation window, τ_{max24} is under-predicted as a result of the grid size being too large to capture the small fluctuations in the flow.

It can be reasoned, then, that the viscous shear stress calculated using a 24×24 pixel interrogation window size for the anatomical measurements would also be underestimated. To correct for this, a corrective mapping algorithm was derived from the results in the axisymmetric model.

The quotient of τ_{max12} and τ_{max24} revealed that the differences were largest for lower values of τ_{max} and smallest for higher values of τ_{max} ; *i.e.*, the scaling factor was not constant. The quotients were sorted into bins corresponding to values of τ_{max24} , and an average scaling factor (SF) for each range of τ_{max24} was calculated; this has been plotted in figure E.2 (c). A polynomial fitted to the result was used to determine the appropriate scaling factors for values of τ_{max24} . As a test, it was

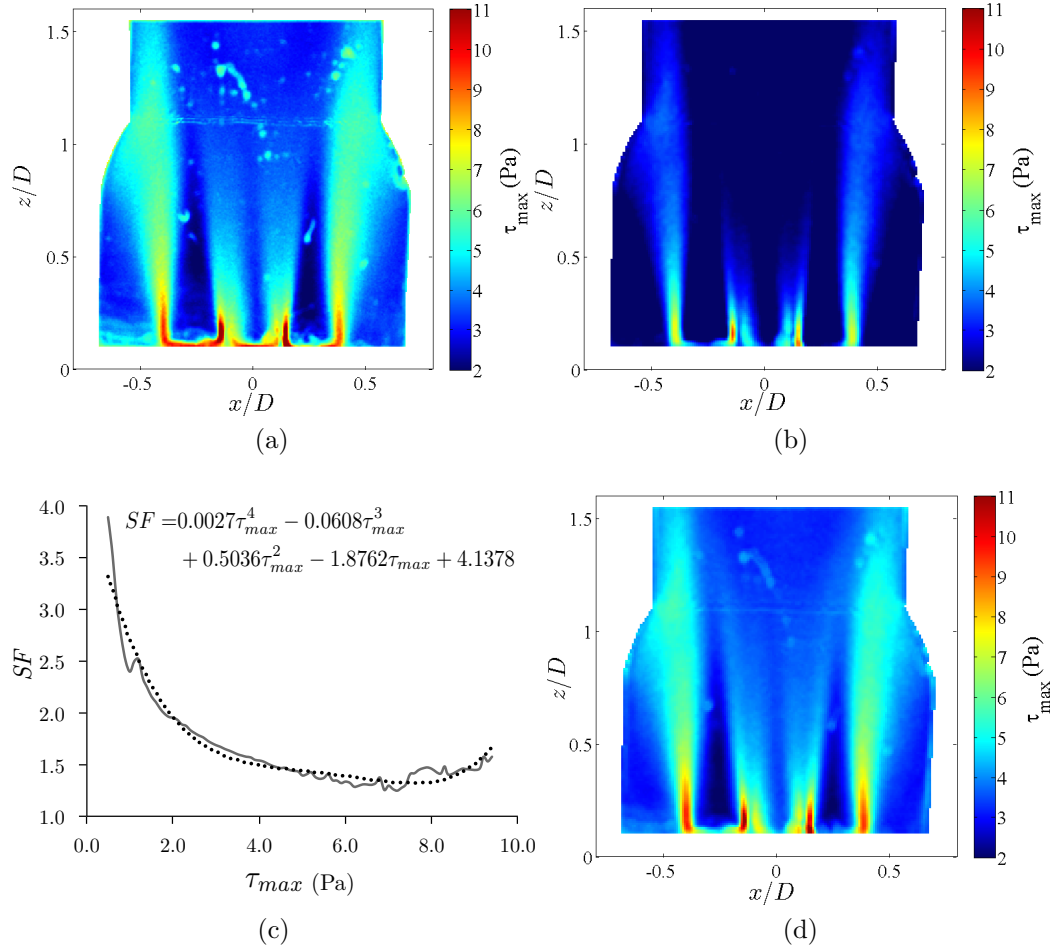


Figure E.2: Viscous shear stress mapping algorithm: (a) τ_{max12} calculated from 12×12 pixel interrogation windows; (b) τ_{max24} calculated from 24×24 pixel interrogation windows; (c) Scaling factor (SF) *vs.* τ_{max24} and fitted polynomial; (d) corrected τ_{max} mapped from τ_{max24} .

applied to τ_{max24} from the axisymmetric model to obtain a corrected measure of τ_{max} (Figure E.2 (d)). This corrected shear stress, mapped from the larger grid data, is nearly identical to that calculated directly from the finer grid. This polynomial was used to correct measurements made in the anatomical model processed with a 24×24 pixel interrogation window.

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