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Shear Stress Distribution and Hemolysis Measurements in a Centrifugal Blood Pump

Matthew Nicholson Giarra

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Shear Stress Distribution and Hemolysis Measurements in a Centrifugal Blood Pump
Shear Stress Distribution and Hemolysis Measurements in a Centrifugal Blood Pump

By

Matthew Nicholson Giarra

A Thesis Presented in Partial Fulfillment of the Requirements for
the Degree of Master of Science in Mechanical Engineering

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Abstract

The use of computational fluid dynamics (CFD) to predict red blood cell trauma (hemolysis) in blood pumps based on their exposure to turbulent stresses has increased in recent years. The U.S. Food and Drug Administration (FDA) has initiated a project to determine the fidelity with which modern CFD can accurately predict hemolysis in such devices. The project involves the collection of experimental data against which externally-conducted CFD simulations may be compared. Because the data will be used to judge the ability of CFD to predict hemolysis, the pump was designed to cause more turbulence and blood damage than would be typical of an approved clinical device. In support of this effort, a shaft-driven centrifugal blood pump was constructed for use in both quantitative flow visualization analysis and in blood-damage experiments. The hydraulic performance of the pump was measured to determine the degree to which it represented a typical blood pump. Particle image velocimetry (PIV) was used to measure planar velocity fields in three different regions of the pump including the blade passage, impeller rear-gap, and cutwater region. For all PIV experiments, the pump delivered volumetric flow rates of 0.6, 3.0, and 6.0 liters per minute (LPM), each at a constant shaft speed of 2800 RPM. Statistical analysis was performed on each PIV data set in order to determine the time-averaged velocity fields as well as to resolve turbulent quantities of interest to the prediction of hemolysis (namely the Reynolds shear stresses). Additionally, the pump was operated using bovine blood as the working fluid in order to measure the hemolysis caused at the same operating points measured during PIV experiments. Further experiments were conducted to determine the contribution of the pump’s shaft-seal interface to the total measured hemolysis. The pump’s hydrodynamic performance was measured to be a close match to that of a typical clinical blood pump. PIV analysis revealed that the velocity and shear stress fields within the pump were dependent on its operating point, and can thus serve as benchmarking data against which to compare CFD analyses. Finally, the pump was confirmed to produce measurable hemolysis. The contribution of a polyurethane shaft seal to the measured hemolysis was significant (39%-62% of the total VAD hemolysis), but this contribution was small (7%-9% of the total VAD hemolysis) when a Teflon seal was used.
Acknowledgements

I couldn’t have started or finished this thesis without the support of a few notable individuals. My experience as a graduate student has been defined by my relationship with my academic advisor, Dr. Steven Day, who presented me with the opportunity to undertake this research and offered me the guidance to complete it. This project would not have been possible without the help of our collaborators at the U.S. Food and Drug Administration. Dr. Richard Malinauskas invited me to perform experiments at the FDA lab, where he and Dr. Luke Herbertson generously sacrificed many late hours to help me collect valuable data. Dr. Prasanna Hariharan was a crucial link between RIT and the FDA, and always made himself available to enthusiastically help solve the various mysteries that invariably surfaced during the course of our research. Rob Kraynik, our department’s mechanical technician, is truly a wizard in the machine shop, and has shown me what great things occur at the intersection of patience, practice, and preparation. Dr. Mark Olles, a fellow rocket enthusiast, has shown me that the pride taken in one’s work is as important to a project’s functionality as machining tolerances and material selection. Dave Gomez, my friend and fellow graduate student whose talent is matched only by his humility, has donated his own share of sweat to the success of this research. Jim Cezo and I began as freshman-year room-mates and rose through graduate school as great friends; his honest advice and practical wisdom was as indispensable to me as any piece of lab equipment. Robert Geisler, my friend and mentor, taught me the importance of doing what you love, or, as he would put it, to “follow your bliss.” Finally, my parents, Jeanna and Paul, are the kindest people I know, and have contributed more to my success than I could ever fit in a one-page acknowledgements section.
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List of Symbols

**Acronyms**
- CFD: Computational Fluid Dynamics
- CHD: Coronary Heart Disease
- CPI: Critical Path Initiative
- FDA: U.S. Food and Drug Administration
- LPM: Liters per minute
- MCSS: Mechanical Circulatory Support System
- Nal: Sodium Iodide
- NIH: Normalized index of hemolysis
- PIV: Particle Image Velocimetry
- PRSS: Principal Reynolds shear stress
- PRNS: Principal Reynolds normal stress
- RBC: Red Blood Cell
- RPM: Revolutions per minute
- VAD: Ventricular Assist Device

**Roman Letters**
- $A_p$: Optical absorbance
- $t_b$: Boundary layer thickness
- $S_{ij}$: Rate of strain due to mean velocities, $ij$ component ($i,j = 1,2,3$)
- $u_i$: Instantaneously measured velocity, $i$ component ($i = 1,2,3$)
- $u_r$: Radial component of velocity
- $u_{th}$: Tangential component of velocity
- $x_i$: $i$ component of position vector ($i = 1,2,3$)
- $\xi$: Characteristic length scale
- $\mathbb{R}$: Set of real numbers
- $C$: Cross correlation
- $D$: Diameter
- $Ht$: Hematocrit
- $I$: Intensity
- $N$: Number of measurements
- $\text{PRNS}$: Principal Reynolds normal stress
Principal Reynolds shear stress
Volumetric flow rate
Radius (constant value)
Viscous shear stress
Variance of relevant variable $x$
Free plasma hemoglobin concentration
Acceleration due to gravity
Turbulent kinetic energy
Hydrostatic pressure
Radius (variable)
Time
Velocity vector
Position vector
Generic time variable for cross correlations
Volume
Kolmogorov (dissipative) time scale
Characteristic velocity scale
Kolmogorov (dissipative) velocity scale
Stress tensor $(i,j = 1,2,3)$
Reynolds stress, $ij$ component $(i,j = 1,2,3)$
Change
Turbulent energy dissipation rate
Kolmogorov (dissipative) length scale
Dynamic viscosity
Kinematic viscosity
Density
Flow coefficient
Angular velocity
Head coefficient
**Other Symbols**

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<tr>
<td>\sim</td>
<td>Of order</td>
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<tr>
<td>' (prime)</td>
<td>Fluctuating (perturbation) component</td>
</tr>
<tr>
<td>\overline{\text{over bar}}</td>
<td>Ensemble averaged value</td>
</tr>
<tr>
<td>\epsilon</td>
<td>Element of</td>
</tr>
<tr>
<td>\nabla</td>
<td>Gradient vector</td>
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<tr>
<td>\partial</td>
<td>Partial derivative with respect to relevant variable $\mathcal{X}$</td>
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Chapter 1: Introduction

Motivations for Research

Coronary heart disease (CHD) is the primary cause of death in the United States, claiming roughly 600,000 lives annually (American Heart Association 2009). While the most effective treatment of CHD is heart transplantation, the relative shortage of donors leaves nearly 90% of patients to seek other options (Mihaylov 2000). In response to this morbid statistic, the clinical use of mechanical circulatory support systems (MCSS) has steadily increased since the first implantation of an artificial heart in 1958 (Akutsu and Kolff 1958). MCSS take the form of specially designed blood pumps that replace or assist the native heart and deliver sufficient blood flow to maintain organ function. The most severe cases of CHD require excision of the native heart and the implantation of a total artificial heart, which assumes complete circulatory responsibility. Less severe CHD may involve some degree of failure of one or both ventricles (the chambers of the heart which contract to pump blood through the circulatory system) in a heart which is still able to provide some sub-optimal function. Such cases often call for the implantation of one or two ventricular assist devices (VADs), which operate in parallel with the native heart. These pumps draw blood from the ventricles and pump it into the aorta (from the left ventricle) or into the pulmonary artery (from the right ventricle). This relieves the unhealthy heart of some of its burden and maintains the patient until a transplant is available, known as "bridge to transplant". In some cases, the added support aids in the recovery of some or all of the native heart's function. In this capacity, VAD therapy is used as a "bridge to recovery." A typical modern VAD is shown implanted in Figure 1.
VADs may operate intracorporeally or extracorporeally, and deliver flow by either a pulsatile diaphragm or a continuous speed rotating impeller. Many early VADs were extracorporeal pneumatically driven pulsatile-flow pumps, which significantly detract from the patient’s quality of life (many are bed-ridden). However, the ever increasing storage capacity of batteries and the decreasing power consumption of modern electronics have given rise to small, portable, and reliable intracorporeal rotary VADs. These VADs operate using batteries similar to those found in laptop computers, and transmit power through a single small transcutaneous cable. The only obvious indications of circulatory support are the battery and pump controller, which are worn on the hip in a small holster. These systems allow unprecedented mobility and a quality of life that closely resembles one allowed by a healthy heart.

Before any VAD is made available for clinical use in the United States, it must apply for and obtain approval of the U.S. Food and Drug Administration (FDA), which is a lengthy and costly process. One of the metrics by which the FDA judges a VAD is the rate at which it...
causes the destruction of red blood cells (hemolysis). The literature indicates a strong correlation between hemolysis and the exposure of RBCs to shear and turbulent stresses (Jones 1995). As such, a first step to predicting the hemolysis caused by a VAD (or any blood-contacting device, for that matter) is to determine the shear stresses it produces. Because the production of shear stress within a device is a fluid-mechanical phenomena, engineers commonly employ computational fluid dynamics (CFD) software to aid in such predictions (Apel 2001), (Song, Throckmorton and Wood, et al. 2003). Next, some experimentally measured threshold time-exposure to shear stress for the generation of hemolysis (known as the hemolytic threshold) is applied to the results of these CFD predictions. This apparently straightforward methodology, illustrated in Figure 2, is complicated by two main issues.

First, the ability of CFD to consistently and accurately predict the behavior of complex flow fields is the subject of some debate. The National Renewable Energy Laboratory recently conducted a "blind comparison" between experimental data and CFD predictions concerning wind-turbine flows, and reported poor correlation with experiments and significant variation among predictions (Simms, et al. 2001). The FDA has discovered similar discrepancies between CFD and experiments in a simple converging-diverging geometry, as shown in Figure 3 (Stewart, et al. 2009). Second, hemolytic thresholds reported in the literature are notoriously inconsistent and vary by several orders of magnitude, as shown in Table 1. These discrepancies call into question the degree to which engineers are able to predict a priori a blood pump's potential to cause hemolysis. As the regulatory body governing the
approval of bio-devices in the United States, the FDA has expressed the need to determine the present ability of CFD to predict hemolysis within blood pumps. This thesis is concerned with the ability of CFD to predict shear-stresses in a blood pump. It should be noted that, while both interesting and important, the determination of the causes of the scatter in hemolytic threshold data in the literature is beyond the scope of the present thesis, and is not considered further. The objectives of the current thesis, then, are as follows:

1. Design and manufacture a prototypical centrifugal blood pump throughout which the velocity and shear-stress distributions may be experimentally measured and which may be used in the present FDA study;
2. Experimentally measure velocity and shear-stress fields within the pump;
3. Conduct preliminary experiments to measure the hemolysis caused by the pump.

Figure 3: Computational and experimental data showing centerline velocity vs axial position in an axisymmetric geometry (Stewart, et al. 2009). CFD results are shown as lines and experimental results as points. The variation of these CFD results illustrates a major obstacle to the consistently accurate prediction of hemolysis within blood pumps, and represents the primary focus of the present thesis.
Table 1: Hemolytic thresholds reported in the literature noting critical shear stresses (viscous and turbulent) and exposure times to cause blood damage. The scatter covers several orders of magnitude in both exposure time and shear stress. While both interesting and important, the determination of the causes of this scatter was beyond the scope of the present research.

U.S. Food and Drug Administration Critical Path Initiative

In March of 2004, in response to a puzzling decline of innovation in the medical industry, the U.S. Food and Drug Administration (FDA) launched the Critical Path Initiative (CPI), an ambitious program to comprehensively modernize the development and testing of biomedical products. This massive undertaking found direction in the Critical Path Opportunities List, an extensive document identifying seventy-six important areas of research that offered immediate promise to the biomedical sciences. Included in the outline, under the heading Manufacturing Devices, was the area of research with which the proposed work is concerned:

"Device Interaction with Blood Flow: Better predictive modeling of the shearing forces and rate of thrombosis caused by implanted devices would enable innovation in physical design and materials."
The current FDA project through which the present thesis was supported, Standardization of Computational Fluid Dynamic Techniques Used to Evaluate Performance and Blood Damage Safety in Medical Devices, was initiated in this vein; it aims to determine the state-of-the-art of computational fluid dynamics (CFD) in relationship to blood flows, particularly concerning hemolysis and thrombosis (clot formation) within biological devices. Interest in participation has been expressed by roughly one hundred individuals affiliated with medical device manufacturers and laboratories in the United States, Germany, France, Australia, Japan, Brazil, and Israel. The project website is available in the Works Cited section of this document.

Relevance of Proposed Work to FDA CPI

As noted previously, CFD simulations are often used in the prediction of hemolysis in blood pumps. Through a better understanding of the capabilities and limitations of modern commercial CFD codes, the FDA may more confidently assess the claims of predicted biocompatibility in device proposals. To this end, several well-defined physical models were created to reproduce important and typical flow regimes that occur in medical devices. These models were the subjects of experiments against which computational models will be evaluated in blind CFD trials and to serve as publicly available benchmarks for future CFD evaluations.

Two different physical archetypes were selected for the experimental portion of the FDA CPI project: a relatively simple "passive" model, consisting of pipe flow through several different transitions in cross-sectional area; and a more complex "dynamic" model of a blood pump representing the many centrifugal VADs currently in use. While the hemodynamics of many commercially available blood pumps have been extensively characterized by their manufacturers, the release of these designs to the general public is infeasible. Because the objectives of the FDA CPI project require unrestricted public access to the model VAD geometry, there exists a need for the design of a novel representative blood pump. The proposed thesis is concerned with the design and characterization of this pump. Because the design of such a device involves a significant level of effort, this thesis is central to the CPI project. The importance of this work and the complexity and importance of predicting blood damage within the pump are illustrated by the devotion of the second blind CFD study to the pump model, as shown in Figure 4.
Figure 4: FDA CPI timeline highlighting aspects with which the present thesis is concerned. "Round-robin" simply means a series of blind trials executed independently by different participants.

**Novelty**

A survey of the literature has found reports of several turbo machinery flow visualization experiments (Wernet 2000), (Day and McDaniel 2005). Additionally, CFD has been used to track the shear-exposure history of computational particles in blood pumps (Song, Throckmorton and Wood, et al. 2003). However, no experimental data regarding blood damage was available for these pumps, and their results were compared to results of blood damage experiments from other pumps. To the knowledge of this author, the proposed thesis will be the first experiment to obtain empirical data for both a pump’s flow field and its blood damage characteristics. Furthermore, the proposed analysis method requires an electronic record of the velocity fields within the entire pump, which is not easily obtainable from previous work in the literature. It is thus more reasonable to perform both flow visualization and blood damage experiments on a pump of novel design than to try and reconstruct data from the literature.
Outline of Thesis

The scope of the present thesis was to perform quantitative flow measurements and preliminary blood-damage experiments in a model blood pump of novel design. Chapter 2 presents background information regarding blood pumps, turbulence, blood-damage, and the experimental technique used for flow measurements (particle image velocimetry). Chapter 3 provides a description of the model pump design including design objectives, analysis, and an overview of the final design. Chapter 4 describes the specific methods by which the present research was performed, including measurement of the pump’s performance, the quantitative flow measurement setup, data processing techniques, and the blood-damage experimental setup. Chapter 5 presents the results of these experiments. Chapter 6 discusses the implications of the data, compares the thesis objectives to the completed work, and recommends improvements for future work. The appendices provide supplementary information regarding the pump design, engineering drawings of custom pump components, and data analysis codes.
Chapter 2: Background

Overview

Blood-contacting medical devices such as catheters, artificial heart valves, and blood pumps are distinguished from the realm of traditional fluid systems by the important consideration of biocompatibility. Of particular concern to the design of these devices is the potential for adverse physical effects due to physiological phenomena associated with the damage of red blood cells (RBCs). Blood damage may take the form of the rupture of RBCs (hemolysis) or of the formation of blood clots (thrombosis) (Turgeon 2004). The present thesis is concerned with hemolysis, which is the failure of the RBC cell wall and is usually caused in medical devices by over-exposure to shear-stress fields. The fluid stresses leading to hemolysis may arise independently from either laminar or turbulent flow phenomena. Jones reports that turbulent stresses are often reported as an assessment of a device's potential to cause hemolysis (Jones 1995). In order to experimentally measure these stresses in the prototype VAD and to relate them to hemolysis, an understanding must be developed regarding (a) the theory of turbulent flows, (b) the relationship of turbulence to hemolysis, and (c) the theory and operation of quantitative flow-measurement techniques. This chapter is meant to provide the reader with some background regarding each of these topics, and may be unnecessary for those already familiar with the material.

Turbulence

Origins of Turbulence

Turbulent fluid flows are characterized by rapid, disorderly fluctuations of velocity and pressure in both space and time (White 1991). This is in contrast to laminar flows, which are smooth and orderly. A laminar flow may become turbulent when it is "tripped," or exposed to some flow perturbation (as caused, for example, by the dimples on a golf ball), and will remain turbulent if some critical value of a flow parameter, such as the Reynolds number, is exceeded. An example of a flow in transition from laminar to turbulent flow over a flat plate is shown in Figure 5. An instantaneous measure of the velocity of a turbulent flow at a point yields the sum of the mean flow velocities and these fluctuating velocity components, or

\[ u_i = \bar{u}_i + u_i' \]  

(1)
where $u_i$ is the instantaneously measured velocity in the $i$ direction, $\bar{u}_i$ is the mean (ensemble-averaged) velocity, and $u_i'$ is the fluctuating component of the $u_i$ component of velocity ($i = 1,2,3$).

Figure 5: Transition from laminar to turbulent flow over a flat plate. Transition occurs, in this case, around $Re = 10^5$ (White 1991)

Application of the Navier-Stokes equations to a turbulent flow gives rise to quantities known as the Reynolds stresses, which are commonly referenced as a measure of a flow's potential for turbulence-induced blood damage. Interestingly, the Reynolds stresses are not actually stresses, but rather are inertia terms that may be mathematically cast as stresses. To show this, it is first noted that the equations of continuity hold for all velocity terms, namely

$$\nabla \cdot \mathbf{V} = 0$$

$$\nabla \cdot \mathbf{V}' = 0$$

Additionally, invoking continuity, conservation of momentum may be expressed as:
Taking the time-average of both sides of this equation can be shown to yield

\[ \rho \frac{DV}{Dt} + \rho \frac{\partial}{\partial x_i} \left( \overline{u_i'u_j'} \right) = \rho g - \nabla p + \mu \nabla^2 \overline{V} \quad (i,j = 1,2,3) \] (4)

The tensor \( \rho \frac{\partial}{\partial x_i} \left( \overline{u_i'u_j'} \right) \) is one of inertia due to fluctuating velocity. However, the above equation is more commonly expressed as

\[ \rho \frac{DV}{Dt} = \rho g - \nabla p + \nabla \cdot \left( \mu \left( \frac{\partial u_i}{\partial x_j} + \frac{\partial u_j}{\partial x_i} \right) - \rho \overline{u_i'u_j'} \right) \quad (i,j = 1,2,3) \] (5)

Or more concisely,

\[ \rho \frac{DV}{Dt} = \rho g - \nabla p + \nabla \cdot \sigma_{ij} \quad (i,j = 1,2,3) \] (6)

Here, the turbulent inertia term has been cast as part of the stress tensor \( \sigma_{ij} \). It is for this reason that the tensor \(-\rho \overline{u_i'u_j'}\) is commonly known as the Reynolds stress, which is denoted by \( \tau_{ij} \). Thus, the nine components of the Reynolds stress tensor are expressed by

\[ \tau_{ij} = -\rho \begin{bmatrix} \overline{u_1^2} & \overline{u_1 u_2} & \overline{u_1 u_3} \\ \overline{u_2 u_1} & \overline{u_2^2} & \overline{u_2 u_3} \\ \overline{u_3 u_1} & \overline{u_3 u_2} & \overline{u_3^2} \end{bmatrix} \] (7)

Note that \( \tau_{ij} \) is symmetric, so that \( \tau_{ij} = \tau_{ji} \).
The Structure of Turbulence

Fluctuating velocities that characterize a turbulent flow take the physical form of "patches" of swirling fluid with randomly oriented velocity components superimposed upon the mean flow. These structures are known as eddies and exist over a range of sizes within a turbulent flow, as shown graphically in Figure 6. Large eddies are responsible for the generation of turbulent kinetic energy within a flow, while small-scale (dissipative) eddies dissipate this energy at the same rate through viscous losses (heat). The smallest scales of turbulence in a flow are known as the Kolmogorov (or dissipative) microscales of length, time, and velocity, defined respectively as

$$\eta \equiv \left( \frac{\nu}{\epsilon} \right)^{1/4}, \zeta \equiv \left( \frac{\nu}{\epsilon} \right)^{1/2}, \nu \equiv (\nu_0)^{1/4}$$

Here, $\nu$ is the fluid's kinematic viscosity and $\epsilon$ is the rate of dissipation of turbulent energy per unit mass. Taylor (Taylor 1935) estimates that

$$\epsilon = \frac{v^3}{\ell}$$

where $v$ and $\ell$ are the characteristic velocity magnitude and length, respectively. Kresta and Wood have estimated the characteristic length scale in stirred-tank experiments as half of the impeller diameter because it is close to the size of the trailing vortices (Kresta 1993). This approximation was applied to the pump, yielding a characteristic length scale $\ell = 25\text{mm}$. The characteristic velocity was taken as the impeller tangential speed at half of the chord length, which gives $v = 5.5 \text{ m/s}$ for a shaft speed of 2000 RPM. Thus, the rate of dissipation of turbulent energy is estimated to be $2 \times 10^{-4} \text{ m}^3\text{ s}^{-1}$. Given the average fluid properties of blood ($\rho \equiv 1000 \text{ kg m}^{-3}, \mu \equiv 5 \text{ mPa s}$), the dissipative length scale is estimated to be about $10\mu\text{m}$. For turbulent flows in biomedical applications, the literature reports dissipative length scales of about 1-10$\mu\text{m}$, which is the size order of a human red blood cell (Jones 1995), and suggests that the present Kolmogorov scale estimation is reasonable.
Figure 6: Eddies of velocity $u$ and size $l$ in a turbulent flow of mean velocity $U$ and boundary layer thickness $L_t$ (Tennekes and Lumley 1972)

**Turbulent Energy Spectrum**

Energy tends to cascade through a continuous spectrum from larger to smaller scale eddies until it is lost to viscous dissipation at the Kolmogorov scale $\eta$, as illustrated in Figure 7. Most of the energy transferred to an eddy is drawn from those a few times larger than itself (Tennekes and Lumley 1972). An understanding of the relationship between large and small-scale turbulence is important for the prediction of hemolysis with PIV. The resolution of PIV is limited by the physical size of particles, the finite size of interrogation regions, and by the resolution of the camera. Small-scale turbulent features often are beyond the resolution of PIV systems. Unfortunately, these features are of great concern to the study of turbulence-induced blood damage. It is therefore desirable to infer some information about small-scale turbulence from the measurement of large-scale turbulence. The capture of 65% of turbulent kinetic energy of the flow has been shown to require spatial resolution of at least $90\eta$. To capture 90% of the turbulent kinetic energy, the spatial resolution must be at least $20\eta$ (Saarenrinne, Piirto and Hannu 2001). In the case of the present work, $90\eta$ and $20\eta$ are roughly 0.8 mm and 0.2 mm, respectively. The PIV system used has captured images with resolution of $15 - 25 \mu m/pixel$ and interrogation regions of $8-16$ pixels, which yields interrogation regions of $120 \mu m - 400 \mu m$. This exceeds the resolution of $90\eta$ in all cases and $20\eta$ in some cases.
Eddies depend on $\nu$ and $u$

Energy cascade

Inertial subrange

Dissipation of energy at rate $\epsilon$

Figure 7: Representation of the cascade of turbulent energy from larger to smaller eddies and dissipating at the Kolmogorov scale (Davidson 2004).

Hemolysis

Hemolysis is the rupture of red blood cells following the failure of the cell wall (Turgeon 2004). Hemolysis may be induced chemically (by exposure to some lysing agent) or mechanically (through exposure to laminar and turbulent fluid stresses). Mechanically-induced hemolysis is an important concern for the designers of blood-handling devices such as artificial heart-valves and VADs (Li, Lo and Lu 2009). While viscous stresses must be considered, they are often much smaller than the Reynolds stresses observed in turbulent flows (Sallam 1982). Several authors have reported threshold Reynolds stresses for hemolysis, as shown previously in Table 1. A size relationship between turbulence scale and hemolysis is shown qualitatively in Figure 8. The literature suggests that RBCs may be damaged by turbulence whose length scales are similar in size to an RBC, but may tolerate larger eddies (Jones 1995). As previously noted, the size of an RBC corresponds to the Kolmogorov scale in typical blood pump flows.
Figure 8: Representation of large scale and Kolmogorov-scale eddies. Large scale eddies, which do not harm blood cells, are several orders of magnitude larger than the Kolmogorov scale eddies, which may lead to hemolysis.

**Measurement of Hemolysis**

When an RBC is destroyed, its internal contents are released into the surrounding fluid which is known as plasma. Included in these cell contents is the protein hemoglobin, which is responsible for oxygen transport. The relative amount of hemolysis in a blood sample may be inferred by measuring the concentration of hemoglobin released into the plasma (known as the free plasma hemoglobin, or \( f_{ph} \)). The increase of hemoglobin in a sample of plasma can be observed qualitatively as a color change from gold to red, as shown in Figure 9.
A quantitative assessment of \( f\text{ph} \) concentration may be performed using optical absorption spectroscopy, which capitalizes on the phenomenon wherein different molecules absorb specific wavelengths of light. Hemoglobin exhibits a strong absorbance peak at 576.5 nm. Cripps determined a relationship between the strength of this peak relative to troughs at 560 nm and 593 nm and the concentration of hemoglobin in a plasma sample (Cripps 1968). To illustrate this technique, the optical absorbance spectra of two samples of plasma containing different \( f\text{ph} \) concentrations are shown in Figure 10. Absorbance \( A_p \) (sometimes known as optical density) is defined as the negative logarithmic ratio between the incident and transmitted light intensities, or

\[
A_p = -\log_{10} \left( \frac{I}{I_0} \right)
\]

Here, \( I_0 \) and \( I \) are the intensities of the light incident to and transmitted through a sample, respectively. Absorbance, naturally, is a unitless number.
Particle Image Velocimetry (PIV)

PIV is an optical measurement technique used to non-invasively obtain planar velocity fields in fluid flows. The fluid is seeded with small (~10 μm) reflective particles whose motion is tracked between images (this is analogous to inferring the velocity of wind by observing the motion of snowflakes). It is well suited for the investigation of flows into which the introduction of traditional probes is difficult, such as turbo-machinery flows (Wemer 2000). For this reason, PIV was used to quantitatively evaluate the flow within the prototype VAD. The technique is shown graphically in Figure 11, and the steps involved are described below.
Figure 11: Principles of PIV operation showing (top to bottom) the generation of and illumination of a test subject a planar sheet and the subsequent capture of images of the flow-field; processing the image pairs to extract velocity vectors; and typical PIV results showing contours and streamlines of velocity in a pump.
Fluid Selection

Because cameras are used to obtain PIV data, optical access to the flow is requisite to any such experiment. When accessing the flow through curved observation windows, the refractive index of the working fluid must often be matched to that of the material of the window to prevent optical distortion, as illustrated in Figure 12 and in Figure 13, which show a scale inserted into an acrylic model whose internal wall is curved. Figure 12 shows the scale in air, whose refractive index is a poor match to that of acrylic. Figure 13 shows the same scale in the same model filled with a liquid whose refractive index is a good match to that of acrylic.

Figure 12: Acrylic model filled with air

Figure 13: Acrylic model filled with refractive index-matched liquid
Particle Seeding

The working fluid in a PIV experiment is seeded with small particles that scatter incident laser light into the camera. The illumination of these particles provides uniformly dispersed fields of high-contrast images that closely follow the flow. The particles are assumed to perfectly follow the flow, and the fluid velocity field may then be inferred by observing the velocities of the particles.

Image Capture

A high-speed digital video camera is focused on a region of interest within the flow. The capture of a pair of images is synchronized with two sequentially fired laser pulses separated by a known time $\Delta t$. The laser acts as a strobe (it provides the "camera flash") whose light is scattered from the seed particles into the camera. The two laser pulses yield a pair of photographs of the field of particles separated by the same time $\Delta t$. One such image is shown in Figure 14.

Figure 14: Photograph of a uniformly seeded flow illuminated by a planar laser
Image Processing

A pair of photographs is broken up into small (i.e., 15-30 pixels) rectangular interrogation regions (images) for analysis. For each pair of images, a velocity vector

\[ \mathbf{V} = \frac{\Delta \mathbf{X}}{\Delta t} = \frac{\mathbf{X}_1 - \mathbf{X}_0}{\Delta t} \]

is sought, where \( \mathbf{X}_0 \) and \( \mathbf{X}_1 \) are the respective positions of the particles in the first and second image and \( \Delta t \) is the time separation between photographs. The pair of interrogation regions are contained as intensity data in discrete arrays \( \mathbf{A} \) and \( \mathbf{B} \), where

\[ \mathbf{A} = f(\mathbf{X}_0, t_0) \quad \text{and} \quad \mathbf{B} = g(\mathbf{X}_0 + \Delta \mathbf{X}, t_0 + \Delta t). \]

Choosing \( t_0 \) and \( \mathbf{X}_0 \) as references, and having set \( \Delta t \) as an experimental parameter, the only unknown in the definition of \( \mathbf{V} \) is \( \Delta \mathbf{X} \). To determine the value of \( \Delta \mathbf{X} \), a cross-correlation \( C \) is performed between \( \mathbf{B} \) and the signal \( \mathbf{A} \) shifted through a range of displacements denoted by the generic variable \( \mathbf{T} \). \( \Delta \mathbf{X} \), then, is taken as the value of \( \mathbf{T} \) at which the maximum of \( C(\mathbf{T}_1, \mathbf{T}_2) \) occurs. An example of a 1-D cross correlation is shown in Figure 15. (Westerweel 1997)

In order to obtain two-dimensional displacements, this same principle may be applied to 2-D data. The discrete 2-D cross-correlation of arrays \( \mathbf{A} \) and \( \mathbf{B} \) of dimensions \((m_a, n_a)\) and \((m_b, n_b)\), respectively, is given by

\[
C(t_1, t_2) = \sum_{m=0}^{(m_a-1)} \sum_{n=0}^{(n_a-1)} A(m, n) \cdot \text{conj}(B(m + t_1, n + t_2)) \quad (11)
\]

where \( 0 \leq t_1 < m_a - 1 \) and \( 0 \leq t_2 < n_a - 1 \). Because \( \mathbf{B} \in \mathbb{R}^{n_a} \) (as \( \mathbf{A} \) and \( \mathbf{B} \) contain only positive integers representing intensity), \( \text{conj}(\mathbf{B}) = \mathbf{B} \), and (11) becomes

\[
C(t_1, t_2) = \sum_{m=0}^{(m_a-1)} \sum_{n=0}^{(n_a-1)} A(m, n) \cdot B(m + t_1, n + t_2) \quad (12)
\]

Thus, the vector \((\mathbf{T}_1)_{m_a} + (\mathbf{T}_2)_{n_a}\) at which the maximum value of \( C \) occurs represents the 2-D displacement (in pixels) from the image \( \mathbf{A} \) captured at \( t_0 \) to the image \( \mathbf{B} \) captured at \( t_0 + \Delta t \). By choosing small interrogation regions, a processed image pair may contain a field of thousands of velocity vectors. A representation of PIV processing is shown in Figure 16.
Figure 15: (a) Two sets of discrete data, $A$ and $B$, where $B$ is a translation and deformation of $A$;

(b) Displacing $A$ by an arbitrary distance $T_1$ (such that the resultant data sets $A(X_0 + T_1)$ and $B(X_0 + \Delta X)$ are dissimilar in phase and form) produces a weak cross-correlation with $B$;

(c) Displacing $A$ by a different distance $T_2$ (such that the resultant data sets $A(X_0 + T_2)$ and $B(X_0 + \Delta X)$ are relatively similar in phase and form) produces a strong cross-correlation with $B$;

(d) Representative relative strengths of discrete cross correlations between $B(X_0 + \Delta X)$ and both $A(X_0 + T_1)$ and $A(X_0 + T_2)$. 

$C_{A,B} = xcorr(A,B)$
Post Processing

In order to obtain statistical information regarding the measured flow, many image pairs (over 1000) are generally captured during each experiment. After velocity vector fields have been obtained for each image pair, statistical quantities such as mean velocity and standard deviation of velocity (which are important to the investigation of turbulence) may be calculated. Additionally, it is during post-processing that the spatial and temporal calibrations are applied to yield data in scientific units. A representative set of post-processed data measured in a turbulent fluid jet is shown in Figure 17. These plots show contours of standard deviation of velocity (indicative of turbulence) as well as streamlines of velocity.
The Measurement of Turbulence Using PIV

As previously noted, Reynolds stresses are often reported as a measure of a turbulent flow's potential to damage blood. As such, it is prudent to measure, if possible, their respective magnitudes in any experiment relating to blood damage. Therefore, the present experiments require development of a method to quantify the Reynolds stresses from the information readily obtained through PIV. While all turbulent flows are three-dimensional, traditional PIV measures only two spatial components of any velocity field, and thus may only be used to obtain part of the total Reynolds stress tensor (stereoscopic PIV can measure all three components of velocity, but was not used). Recalling all the components of the Reynolds stress tensor,

\[
\tau_{ij} = -\rho \left[ \begin{array}{ccc}
    u_i'' u_i'' & u_i'' u_j'' & u_i'' u_k'' \\
    u_i'' u_j'' & u_j'' u_j'' & u_j'' u_k'' \\
    u_i'' u_k'' & u_j'' u_k'' & u_k'' u_k'' \\
\end{array} \right] \tag{13}
\]

Again, \(\tau_{ij}\) is symmetric so that \(\tau_{12} = \tau_{21}\), etc. Also recall that

\[
u_i' = u_i - \bar{u_i} = \frac{\partial x_i}{\partial t} - \frac{\partial x_i}{\partial t} (i = 1,2,3) \tag{14}
\]
For normal PIV operation with a thin laser sheet, $\delta x_3 = 0$. Thus $u_3'$ is unknown and is treated to be near zero. Thus, the measurable components of the Reynolds stress are reduced to

$$
\tau_{ij} = -\rho \begin{bmatrix}
\frac{\partial u_i'}{\partial x_j} & \frac{\partial u_i'}{\partial y_j} \\
\frac{\partial u_j'}{\partial x_i} & \frac{\partial u_j'}{\partial y_i}
\end{bmatrix}
$$

(15)

Also,

$$
\bar{u_i'u_j'} = \frac{1}{N} \sum_{k=1}^{N} (u_i - \bar{u}_i)(u_j - \bar{u}_j) \quad (i, j = 1, 2)
$$

(16)

Here, $N$ is the total number of velocity measurements considered for an experiment. Multiplying by density to obtain the Reynolds stress components,

$$
\tau_{ij} = -\rho \frac{1}{N} \sum_{k=1}^{N} (u_i - \bar{u}_i)(u_j - \bar{u}_j) \quad (i, j = 1, 2)
$$

(17)

When $i = j$,

$$
\tau_{ii} = -\rho \frac{1}{N} \sum_{k=1}^{N} (u_i - \bar{u}_i)^2 \quad (i = 1, 2)
$$

(18)

Note that this equation closely follows the form of statistical variance, given by

$$
\text{Var}(X) = \frac{1}{N} \sum_{k=1}^{N} (X_k - \bar{X})^2
$$

(19)

Recasting $\tau_{11}$ and $\tau_{22}$ in this form, the three independent components of the measurable Reynolds stress tensor are

$$
\tau_{11} = -\rho \cdot \text{Var}(u_1)
$$
Thus, a single PIV measurement may infer only three of six independent Reynolds stress components. This limitation implies that such measurements are accurate only insofar as the out-of-plane velocity is small compared to the in-plane velocities. This requirement is satisfied in the small gap of the blade passage of a centrifugal pump whose impeller height is small compared to its radius. It is also satisfied in the volute, where the mean fluid velocity is very nearly tangent to the volute profile. As such, PIV measurements should yield accurate measurements of Reynolds stress in the blade passage and volute of the proposed pump.

**PIV Error**

The two main sources of error in PIV are the experimental setup and the image processing algorithms (Westerweel 1997). Sources of error caused by the experimental setup include lens/camera aberrations, camera noise, improper particle seeding density, improper time spacing \((\Delta t)\) between images, and large out-of-plane velocities. Sources of processing errors include the implementation of cross-correlation algorithms, displacement peak-finding schemes, and the relative size of interrogation regions to the sizes of particles. Carefully executed PIV experiments can yield velocity measurements accurate to 1% full-scale values.

In order to determine the error due to image-processing algorithms independently from that due to the experimental setup, PIV analysis was performed on an idealized computer-generated pair of photographs. Each photograph consisted of a 1000-by-1000 pixel array of bits representing a periodic arrangement of identical black circles against a white background. Each circle was 60 pixels in diameter. The second photograph ("image B") was formed by shifting each circle from the first photograph ("image A") 50 pixels down and 50 pixels to the right. Each interrogation region was 200 pixels in both dimensions for a total of 25 interrogation regions. Figure 18 shows the two "ideal" images analyzed along with a superimposed grid representing the array of interrogation regions. The images were processed, and the values of the calculated and specified displacements were compared.
all cases, the error between calculated and specified displacements was below 0.02% of the prescribed displacement. The resulting velocity vectors are shown in Figure 19.

Figure 18: Computer-generated image pair used for PIV software error analysis. Each shifted circle was moved 50 pixels down and to the right, and each square represents a single interrogation region. Note that the grid is displayed only to represent the interrogation regions, and was not present in the analyzed images.

Figure 19: Velocity vectors calculated by PIV processing software showing displacement of computer-generated image pair. The error for all vectors was below $2 \times 10^{-5}$ % of the full-scale displacement. The horizontal and vertical axis labels represent the number of pixels in each direction.
Chapter 3: Pump Design

Design Requirements

A prototypical centrifugal blood pump was designed for use in PIV and hemolysis experiments. The FDA working group determined the design requirements, and the decision process leading up to the final design criteria are shown graphically in Appendix A. The pump design was based on the following requirements.

1. Pump performance should be similar to that of typical centrifugal VADs;
2. Pump should contain simple geometry to facilitate computational modeling (i.e., grid generation);
3. Pump should generate appreciable hemolysis to help quantify robustness of CFD predictions;
4. Pump must be optically accessible to allow interrogation by PIV;
5. Pump must allow optical access to gap between rear impeller shroud and lower housing wall;
6. Clearance between impeller blade tips and upper housing should be adjustable;
7. Clearance between rear impeller shroud and lower housing wall should be adjustable;
8. Position of pump relatively to laser sheet and camera field of view should be adjustable.

Hydraulic Design

In order to allow for ease of computational meshing, the prototype VAD was designed with a straight-bladed impeller. A diameter of 52mm was chosen to represent the size of a typical centrifugal VAD and to allow a large portion of the impeller to fit into the camera field of view without an unreasonable sacrifice of image resolution.

In centrifugal pumps, the volute serves to convert the kinetic energy imparted on the fluid by the impeller into hydrostatic pressure. Efficient volute designs often take the form of logarithmic spirals whose radius increases with angular position. Generally, the volute is designed such that the fluid exits the impeller with a tangential velocity equal to that of the blade’s trailing edge, and so that the fluid exits the pump at the designed volumetric flow rate. Such analysis was performed for the selected VAD size and flow rate, and the volute diameter was found to grow by only about 7%, from 58.0 mm to 62.6 mm. Thus, in order to simplify both manufacturing and computational modeling, the volute was specified to have a
constant diameter. According to Gindhar and Moniz, the diameter of an axisymmetric volute should be about 1.15 times that of the impeller (Gindhar 2004). Thus, the FDA VAD was given a volute with a constant diameter of 60mm.

**Design for Hemolysis**

In order to obtain blood-damage data against which CFD predictions may be compared, the FDA VAD was intentionally designed to generate hemolysis at a rate beyond that which would be acceptable in a clinical pump. To this end the Reynolds shear stresses were estimated a priori following Prandtl’s postulation, which is presented by Tennekes and Lumley (Tennekes and Lumley 1972). For the sake of estimation, the turbulence was assumed to be isotropic (i.e., $u_u - u_i = u_j$). Crude as this assumption may be, all that was sought was an order-of-magnitude estimation for the magnitude of the Reynolds shear stresses. According to Prandtl, for such isotropic turbulence,

$$
\tau_{i2} = \rho \varepsilon \left( \frac{\partial u_2}{\partial x_1} \right)^2 \tag{21}
$$

Here, $\varepsilon$ is a scalar constant of order 1. The characteristic turbulence length $\ell$ was taken as size of the gap between the impeller tip and the housing wall, anticipated to be about 2 mm. The gradient of the mean tangential velocity $\left( \frac{\partial u_1}{\partial x_2} \right)$ was taken as the impeller speed at half its chord length over the gap $\ell$. The density $\rho$ was taken as that of blood, or about 1050 $\text{kg/m}^3$.

Recasting the velocity gradient as in this way, (21) can be re-written as

$$
\tau_{i2} = \rho \varepsilon \left( \frac{\omega R}{\ell} \right)^2 \tag{22}
$$

Solving for $\omega$,

$$
\omega = \frac{2}{R} \frac{\tau_{i2}}{\rho \varepsilon} \tag{23}
$$
Thus an impeller speed may be specified based on the desired Reynolds shear stress, fluid density, impeller radius, and a conservative guess at the value of the constant \( c \).

In order to determine the threshold shear stress for hemolysis, an estimate of the residence time of a blood cell within the pump is helpful. To obtain such an estimate, the pump impeller is modeled as a perfect source flow, whose velocity is purely radial. Invoking the definition of radial velocity,

\[
\frac{dr}{dt} = \frac{u_r}{u_r}
\]

or

\[
\frac{dt}{dr} = \frac{u_r}{u_r}
\]

Here, \( r \) is the radial spatial coordinate. Also, the differential volumetric flow rate of fluid exiting the impeller with purely radial velocity is given by

\[
dQ = u_r dA
\]

The radial area \( dA \) is given by

\[
dA = (2\pi h)dr
\]

Here, \( h \) is the impeller height. Thus,

\[
dQ = (2u_r\pi h)dr
\]

Combining like terms and integrating both sides,

\[
\int u_r dr = \int \frac{dQ}{2\pi h}
\]

Solving for \( u_r \) yields
This is the familiar radial velocity of a perfect source or sink. Substituting this term into the definition of radial velocity and integrating both sides,

\[
\int_0^t \frac{dt}{Q} = \int_{r_h}^{R_0} \frac{2 \pi r h}{Q} \, dr
\]

\( R_0 \) and \( R_i \) are the outer and inner radial positions. Performing this integral yields

\[
t = \frac{n h}{Q} \left( R_0^2 - R_i^2 \right)
\]

Specifying the impeller diameter and considering the maximum residence time by taking the limit as \( D_i \to 0 \).

\[
t = \frac{n h D_i^2}{4Q}
\]

From the design of typical blood pumps, \( h \approx 3 \text{ mm}, Q \approx 6 \text{ L/min}, D_0 \approx 5 \text{ cm} \). Thus

\[
t = \frac{\pi (3 \times 10^{-3} \text{ m}) (0.05 \text{ m})^2}{4 \left( \frac{L}{\text{min}} \times \frac{1 \text{ min}}{60 \text{ sec}} \right)}
\]

\( \therefore t \approx 6 \times 10^{-2} \text{ sec} \)

Some literature suggests that exposure times in the neighborhood of \( 10^{-2} \) to \( 10^{-1} \) seconds correspond to critical Reynolds stresses of 200 – 400 Pa (Sutera and Mehrjardi 1975), (Sallam and Hwang 1984). In an attempt to ensure that hemolysis is generated, the turbulence prediction was very conservative, and a Reynolds shear stress of 1000 Pa was specified in the estimation given by (23). Taking the impeller radius as 25 mm, and
conservatively setting the constant \( c \) as 1, the required shaft speed is estimated to be 750 RPM. In a further attempt to ensure the significant generation of hemolysis, this shaft speed was multiplied by 3 for a design speed of about 2300 RPM. Thus, a centrifugal pump with an impeller diameter of 50mm and a shaft speed of 2300 RPM should create sufficient turbulent stresses to cause hemolysis. The maximum pressure rise corresponding to this shaft speed, as predicted by Bernoulli’s principle, is about 135 mmHg, which is reasonable for a centrifugal VAD.

**Optical Access**

A common problem in the visualization of turbo machinery flows is the optical occlusion of different internal regions by opaque boundaries such as impeller blades (Wernet 2000). One side of a blade, for example, generally hides the other side from illumination, resulting in incomplete flow field measurements. While surveying the literature, no experiments were found to have obtained total optical access to a pump. To ensure optical accessibility to all regions of interest in the FDA CPI pump, its internal and external components were constructed from transparent materials (acrylic) wherever practical.

**Final Design**

The final pump design consisted of an impeller assembly, upper and lower housings, a bearing-supported drive shaft, and a mechanical shaft seal. The impeller assembly contained an acrylic 52mm diameter outer “ring” with four straight blades, which was epoxied onto a one-piece stainless steel hub. The hub was polished to reduce the scattering of incident laser light into the camera. Additionally, it contained internal threads to accommodate a 1/8 inch drive shaft. The hub was physically located on the shaft by a precisely machined slip-fit interface before the threads engaged. The drive shaft passed through a pair of 1/8” ID ball bearings and a polyurethane shaft seal. The shaft was polished to be axially free within the bearings and allow for precise positioning of the impeller within the housing. The direction of the impeller’s rotation was selected such that the hub/shaft connection would tend to tighten rather than loosen during operation. The upper housing contained a 60mm diameter cylindrical cavity which served as a constant-radius volute, as well as a 10° half-angle conical diffuser. The housing was polished to reduce optical distortion of the fluid within the pump. Additionally, 12 mm ID stainless steel inlet and outlet tubes were pressed into the upper housing, and were each equipped with a pressure tap made of 20-gage hypodermic tubing. The lower housing contained an elevated platform to separate the impeller from any optically opaque hardware and to accommodate an axial o-
ring, which provided a water-tight seal between the housing halves. The gap between the upper and lower housing halves was adjusted with spacers made from shim-stock, and could be controlled in increments of 0.030 inches (0.76 mm). The housing halves were held together with four screws positioned so as not to obstruct the laser sheet or the capture of images. To eliminate the possibility of damaging the housing by stripping fragile acrylic threads, the screws mated with threaded brass inserts pressed into the lower housing. An exploded view of the prototype VAD is shown in Figure 20. Detail of the impeller assembly is shown in Figure 21. Detailed engineering drawings of all custom pump components may be found in Appendix B.

Figure 20: Exploded view of prototype centrifugal VAD used for PIV and blood damage experiments
The pump assembly was rigidly attached to an optical breadboard (a precisely machined flat plate containing a 2-D array of uniformly spaced threaded holes usually used for the alignment of optics) to facilitate alignment with the drive components and integration into the optical alignment hardware. The impeller was driven by a Faulhaber 4490-B brushless DC motor. The shaft speed was regulated using the Faulhaber MPV hardware and software and verified by observing the frequency output of a U.S. Digital 500 line-per-revolution optical encoder attached to the motor’s drive shaft. The motor was mounted atop a ThorLabs MT-1 single axis translation stage with a range of motion of ¾ inch and precision of 0.001 inch. The pump’s drive shaft was connected to that of the motor with a Stafford Manufacturing rigid shaft coupling. Because the drive shaft was axially unconstrained within its support bearings, an axial translation of the motor resulted in an axial translation of the impeller. This assembly is shown in Figure 22 and in Figure 23.
Shims allow for variable housing gap.

Single-axis translation stage allows for precise axial positioning of impeller within housing.

Figure 22: Rendering of pump and drive/positioning hardware

Figure 23: Side view of pump and drive/positioning hardware
Chapter 4: Methods

An optically accessible prototype pump was constructed for use in both PIV analysis and in blood-damage experiments. The performance of the pump was measured in order to determine the operating points at which its hydraulic performance was representative of that of a typical VAD. PIV was used to measure planar velocity fields in three different regions of the pump including the blade passage, impeller rear-gap, and cutwater region. For each region of interest, the pump delivered volumetric flow rates of 0.6, 3.0, and 6.0 liters per minute (LPM), each at a constant shaft speed of 2800 RPM, for a total of nine sets of PIV data. Statistical analysis was performed on each data set in order to determine the time-averaged velocity fields as well as to resolve turbulent quantities of interest to the prediction of hemolysis (namely the Reynolds shear stresses). Finally, the pump was run using bovine blood as the working fluid in order to measure the hemolysis it causes at the same operating points specified during the PIV experiments.

Working Fluid

As shown in Figure 24, human blood is a non-Newtonian shear-thinning fluid whose kinematic viscosity decreases from 0.04 to 0.08 Stokes from shear rates of 1 — 200 \( \text{s}^{-1} \). (Long 2005). Many modern computational and experimental studies are performed with Newtonian blood analogs (Day and McDaniel 2005), (Song, Throckmorton and Wood, et al. 2003). In keeping with this practice, the working fluid for the present PIV experiments was a Newtonian blood analog consisting of a mixture of distilled water and sodium iodide (NaI) whose refractive index was a close match to that of acrylic. The viscosity and density of the blood analog were measured to be 4.7 mPa•s and 1860 kg/m\(^3\), respectively.
Blood Kinematic Viscosity

Figure 24: Dependence of human blood viscosity on shear rate. While it does exhibit non-Newtonian behavior, the viscosity of blood decreases only slightly with increasing shear rate (Long 2005).

Pump Performance

Pump performance data was collected for pumping water at impeller speeds from 1000-4000 RPM at increments of 1000 RPM. For each speed, volumetric flow rate and pressure rise were recorded at several operating points. Volumetric flow rate was controlled by constricting the pump’s rubber outlet tube with a screw-type clamp, and was measured using an Em-Tec ultrasonic flow meter with a claimed accuracy of ± 7% of the total flow rate from 1-10 LPM. The flow meter was clamped onto the flexible tubing exiting the fluid reservoir. Pressure rise was measured with a Validyne DP-15 differential pressure sensor with a range of 0 – 165 mmHg and a claimed accuracy of ±0.4 mmHg. Pressure taps were made of 20-gage (600 µm ID x 900 µm OD) steel hypodermic tubing mounted flush with the internal surface of the steel inlet and outlet tubing. The data was non-dimensionalized to head and flow coefficients defined respectively by
In order to locate the design operating point (6 LPM at 100 mmHg), the data was re-dimensionalized by estimating pressure rise and flow rate at different impeller speeds for which data was not recorded. A third-order polynomial was fit to the resulting data to generate a continuous pump performance estimate.

**PIV Setup**

The acrylic pump and drive components were rigidly mounted to a 2-axis translation stage on an optical table. Adjustment of these stages allowed precise control of the pump’s position relative to the laser sheet and camera. An optical slide served as a third axis, and allowed for coarse position control. A CAD model of the pump integrated with the PIV positioning hardware is shown in Figure 25, and a photograph of the same setup is shown in Figure 26. Images were captured in three physical regions, referred to as the "blade passage," the "back gap," and the "cutwater region," as shown in Figure 27. A photograph of the PIV experiment showing the camera, pump, and auxiliary equipment is shown in Figure 28. Images were captured with an IDT Motion Pro X-3 Plus high-speed digital video camera equipped with a Nikon lens with focal length 90mm, f-number 2.8, and image reproduction ratio 2:1. The distance from the camera lens to the pump was chosen such that an entire quadrant of the volute fit into the field of view, resulting in spatial resolution of approximately 30μm per pixel. The fluid was illuminated with a New Wave Research Solo II pulsed 532nm laser. A planar laser sheet was created by passing the collimated laser beam through a cylindrical lens of focal length 50mm and a spherical lens of focal length 400mm. The camera was focused on the region of the pump in which the laser sheet’s waist (analogous to the focal point of a beam) occurred, which measured approximately 1mm thick. Image capture was synchronized with the pump impeller’s position using the motor’s shaft encoder, which output a 5-volt pulse once per revolution. The encoder pulse was passed to a Berkley Nucleonics Corporation pulse delay generator, which output a similar 5-volt pulse after a time delay controllable to 1ms. The position of the pump impeller within the image was

\[
\phi = \frac{gh}{(\omega aD)^2} 
\]

\[
\psi = \frac{Q}{aD^2} 
\]
controlled by varying this time delay. The camera, laser, and encoder were synchronized using a TSI Laser Pulse Synchronizer. Images were recorded using TSI Motion Pro software loaded on a standard workstation computer. The synchronization setup is shown schematically in Figure 29.
Figure 26: Completed prototype VAD mounted on translation stages for PIV experiments

Figure 27: Measurement locations in pump. The forward-view of the measurement locations is shown on the left, and axial positions of the laser sheet are shown on the right. The cutwater region and blade passage images were captured in the axially right-most plane, while the back gap images were captured in the axially leftmost plane.
Figure 28: PIV setup showing relative position of camera (left) and pump setup.

Figure 29: Schematic view of the synchronization setup used to set the impeller position at which images were captured.
The pump was operated at 2800 RPM at flow rates of 6.0, 3.0, and 0.6 LPM. 1000 images were captured during each experiment. The flow rate was controlled by restricting the outlet tubing with an adjustable hose clamp and monitored with an ultrasonic flow meter for the duration of each experiment. The shaft speed was monitored via the frequency of the encoder signal. The space between the underside of the rotor and the lower housing wall was maintained at 1mm, which was observed to be the smallest gap that allowed for the capture of clear images in the back-gap region. The separation between housing halves was maintained at 3mm, resulting in a 2mm gap between the impeller blades and the upper housing. For each operating point, data was captured within the blade passage, behind the rotor, and in the cutwater region.

Data Processing

Velocity vectors were obtained for each image pair using WIDIM, a code written by Fulvio Scarano of Delft University of Technology. Each image was resolved into 127 x 79 interrogation regions in the vertical and horizontal image directions, respectively. To spatially calibrate the images for conversion into SI units, the number of pixels constituting a known dimension within the pump (i.e., the distance between a blade’s trailing edge and the volute wall) was measured and assumed to be invariant throughout each data set. Ensemble statistics were performed on each set of vector fields to determine the mean velocity, fluctuating velocity, and number of valid vectors (based on signal/noise ratio as determined by WIDIM) at each point. In order to determine the velocity of the fluid relative to the impeller’s rotating frame of reference, the origin of the image was assumed to coincide with the impeller’s axis of rotation. Velocity vectors were then transformed from the Cartesian to the polar coordinate system by applying the transformation matrix described by

\[
\begin{bmatrix}
U_r \\
U_\theta
\end{bmatrix} =
\begin{bmatrix}
\cos \theta & \sin \theta \\
-\sin \theta & \cos \theta
\end{bmatrix}
\begin{bmatrix}
U_x \\
U_y
\end{bmatrix}
\]

The tangential velocity of the impeller was then subtracted from that of the fluid to obtain relative velocity.

The normal and shear components of the Reynolds stress tensor were calculated based on velocities measured in a Cartesian coordinate system whose vertical and horizontal axes lay along those of the rectangular images collected. It is important to note
that the measured magnitudes of the components the Reynolds stress tensor depend on the orientation of the chosen coordinate system. The Reynolds stresses, however, are physical quantities whose magnitudes, in reality, are invariant with respect to the observer's point of view. The maximum value of each component may be found from the measured components by using the equation given by Baldwin et al (Baldwin 1993) as

\[
PRNS = \frac{T_{11} + T_{22}}{2} + \sqrt{\left(\frac{T_{11} - T_{22}}{2}\right)^2 + T_{12}^2 + T_{21}^2}
\]  
(38)

\[
PRSS = \sqrt{\left(\frac{T_{11} - T_{22}}{2}\right)^2 + T_{12}^2 + T_{21}^2}
\]  
(39)

Here, PRNS and PRSS represent the principal Reynolds normal stress and shear stress, respectively.

**Hemolysis Testing**

As part of the FDA Critical Path Initiatives project, computational modelers will attempt to predict the hemolysis caused by the FDA VAD at various operating points. In order to obtain empirical hemolysis data against which these predictions may be compared, two existing FDA VAD rigs were used to perform hemolysis tests of (1) the entire VAD and (2) the isolated shaft-seal interface. Both sets of tests were performed at a shaft speed of 2800 RPM, which corresponds to the VAD's ideal operating point. Fresh bovine blood (under 48 hours old) was used as the working fluid in all tests. Sodium citrate was added as an anti-coagulant to reduce thrombus formation. The hematocrit ranged from 0.36 – 0.37. Both flow loops were rinsed with phosphate buffered saline prior to each test. Optical spectroscopy was used to measure hemolysis according to the method developed by Cripps (Cripps 1968) and identified as particularly safe, easy, and precise among hemolysis analyses (Malinauskas 1997).
VAD Hemolysis Testing

The VAD was operated at volumetric flow rates of 0.6, 3.0, and 6.0 liters per minute. The testing procedure used was defined by ASTM F 1841 – 97: Standard Practice for Assessment of Hemolysis in Continuous Flow Blood Pumps, with a few exceptions. Most notably, the duration of each test was reduced from 6 hours to 1-2 hours. Additionally, the pressure rise across the pump was not specified so as not to over-define the pump's operation. Instead, the set parameters were shaft speed and volumetric flow rate. The two shaft seals used were made of FDA-compliant polyurethane and of graphite-reinforced Teflon, respectively. The gap between the impeller shroud and the lower pump housing was set to 1 mm, and the gap between the blades and the upper housing wall was 2 mm.

For tests using the polyurethane seal, three replicates each were performed for flow rates of 6.0 and 0.6 LPM, and two replicates were performed for 3.0 LPM. For tests using the Teflon seal, two replicates each were performed for flow rates of 6.0 and 0.6 LPM. Two 2 ml blood samples were collected for analysis at evenly spaced intervals during each test, with the first sample taken at the beginning of each test. The VAD is shown during blood testing in Figure 30. Pump dimensions and other constants are shown in Table 2.
Figure 30: FDA VAD filled with blood during hemolysis testing

<table>
<thead>
<tr>
<th>Total VAD Hemolysis Tests</th>
<th></th>
</tr>
</thead>
<tbody>
<tr>
<td>Impeller Description</td>
<td>4 straight blades</td>
</tr>
<tr>
<td>Impeller diameter (mm)</td>
<td>52</td>
</tr>
<tr>
<td>Blade height (mm)</td>
<td>3</td>
</tr>
<tr>
<td>Blade width (mm)</td>
<td>3</td>
</tr>
<tr>
<td>Shroud thickness (mm)</td>
<td>4</td>
</tr>
<tr>
<td>Rear gap (mm)</td>
<td>1</td>
</tr>
<tr>
<td>Front gap (mm)</td>
<td>2</td>
</tr>
<tr>
<td>Upper / lower housing spacing (mm)</td>
<td>3</td>
</tr>
<tr>
<td>Shaft speed (RPM)</td>
<td>2800</td>
</tr>
</tbody>
</table>

Table 2: FDA VAD information
**Shaft Seal Hemolysis Testing**

The FDA VAD was driven by an external motor coupled to the impeller by a rigid shaft and thus incorporated a shaft seal. Because the dynamics of hemolysis at the shaft-seal interface are not well understood, the seal is unlikely to be modeled in computational experiments during the FDA Critical Path Initiative project. As such, it was necessary to experimentally determine the seal's contribution to the hemolysis measured during pump experiments. To this end, an FDA VAD was modified to isolate the shaft-seal interface for hemolysis testing. Both polyurethane and Teflon seals were tested, and are shown in FIGURE. The impeller was removed from a standard FDA VAD, leaving only the shaft protruding from the seal in the housing. Blood was pumped through the housing at 0.6 LPM using an auxiliary blood pump. In an attempt to minimize the hemolysis due to the flow loop, the VAD outlet served as the seal tester inlet in hopes of maintaining a somewhat orderly flow. In order to distinguish between the hemolysis caused by the shaft - seal interface and that caused by the flow loop, tests were performed with the shaft at rest and with the shaft spinning at 2800 RPM. Each test was run for 1-2 hours, and five 2 mL blood samples were collected at evenly spaced intervals during each test. 2 to 4 replicates of each test (static and spinning) were performed. The hemolysis due to the spinning shaft was taken as the average hemolysis measured during "non-spinning" tests subtracted from that measured during "spinning" tests. The shaft seal test information is summarized in Table 3. The seal tester is shown in Figure 32.
Figure 31: Shaft seals tested in hemolysis experiments included one made of polyurethane (left) and one of graphite-reinforced Teflon (right).

<table>
<thead>
<tr>
<th>Shaft seal tests</th>
<th>Seal 1</th>
<th>Seal 2</th>
</tr>
</thead>
<tbody>
<tr>
<td>Seal supplier</td>
<td>American High Performance Seals</td>
<td>McMaster Carr</td>
</tr>
<tr>
<td>Seal material</td>
<td>Polyurethane</td>
<td>Graphite-reinforced Teflon</td>
</tr>
<tr>
<td>Shaft material</td>
<td>Stainless steel</td>
<td>Stainless steel</td>
</tr>
<tr>
<td>Seal ID (inches) (mm)</td>
<td>0.125 (3.18)</td>
<td>0.125 (3.18)</td>
</tr>
<tr>
<td>Seal OD (inches) (mm)</td>
<td>0.375 (9.51)</td>
<td>0.25 (6.35)</td>
</tr>
<tr>
<td>Shaft speed (RPM)</td>
<td>2800</td>
<td>2800</td>
</tr>
</tbody>
</table>

Table 3: Shaft seal information
Figure 32: Shaft seal tester showing booster pump (on left) delivering blood to shaft-seal test chamber (on right)

**Analysis**

*ASTM F 1841 – 97* cites normalized index of hemolysis (NIH) as the metric by which rates of hemolysis may be compared among blood pumps, where NIH is given by the document as

\[
NIH = \frac{\Delta \text{fph} \times V \times 100 - H_t}{Q \times t}
\]  

(40)

Here,  
\(\Delta \text{fph}\) = increase of plasma free hemoglobin concentration (g/L) over the sampling time interval;  
\(V\) = circuit volume (L);  
\(Q\) = flow rate (LPM);  
\(H_t\) = hematocrit (%);
$t$ = time (from start of test) at which sample was collected (minutes).

Hematocrit $Ht$ is the percent by volume of red blood cells to total blood volume. Due to the nature of the tests, a different metric was deemed necessary to compare the FDA VAD with the shaft seal tester. NIH normalizes hemolysis data with respect to volumetric flow rate, test volume, hematocrit, and time. However, the hemolysis caused by the shaft seal tester is expected to depend on shaft speed only, and not on volumetric flow rate. Thus, the hemolysis measured in each VAD test was compared to that measured during each seal test by normalizing the hemolysis data with respect to initial blood volumes only. Thus, based on the criteria that the hemolysis caused by the seal should be below 10% of that measured in the total VAD, an acceptable seal would yield

$$\Delta f_{ph_{VAD}} \geq 10 \frac{V_{seal\, tester}}{V_{VAD}}$$

(41)

Here, $\Delta f_{ph_{VAD}}$ is the change in hemoglobin concentration observed during the VAD tests; $\Delta f_{ph_{seal\, tester}}$ is the average hemoglobin content measured during each non-spinning test subtracted from the average of that measured during the spinning tests; and $V_{seal\, tester}$ and $V_{VAD}$ are the initial blood volumes contained in the seal tester and VAD flow loops, respectively.
Chapter 5: Results

Pump Performance

The collected data indicates that the pump performance was similar to that of typical centrifugal blood pumps, and is shown dimensionally and non-dimensionally in Figure 33 and in Figure 34, respectively. As expected, the non-dimensionalized data obtained at each impeller speed collapsed into a single curve. The design operating point was thus estimated to occur at 2800 RPM, as shown in Figure 35. Additional pump performance data collected at 2800 RPM using whole bovine blood as the working fluid is shown in Figure 36. Finally, the shaft seal began to leak after roughly 24 hours of PIV experiments. Upon inspection, the shaft exhibited deep gouges at the shaft-seal interface.

![FDA VAD Hydraulic Performance](image)

Figure 33: Measured pump performance at impeller shaft speeds from 1000-4000 RPM.
Figure 34: Non-dimensionalized pump performance. This data may be used to estimate the performance of the pump at any dimensional operating point.

Figure 35: Estimated pressure rise vs. flow rate based on non-dimensional pump analysis (ω = 2800 RPM)
Figure 36: FDA VAD performance data collected using bovine blood as the working fluid. The measured pump performance closely matches that predicted by non-dimensional analysis.

PIV Results

High-contrast, evenly illuminated, and consistently-synchronized PIV images were achieved for all data sets. A sample image captured in the VAD’s cutwater region is shown in Figure 37. A series of images illustrating synchronization consistency is shown in Figure 38. Contour plots showing speed with streamlines, principal Reynolds shear stress magnitude (PRSS), viscous shear stress magnitude (SS), and vector quality (Count) in the blade passage, back gap, and cutwater region at 0.6, 3.0, and 6.0 LPM, along with histograms showing the Reynolds shear stress distribution in each data set, are shown in Figure 39 - Figure 50. The fluid speed relative to the impeller is reported for the blade passage and back-gap regions, while absolute fluid speed is reported in the cutwater region. The data considered in the histograms was taken only within the fluid-containing regions of the PIV images. This was accomplished by masking the images and discarding data that fell outside of the fluid flow (i.e., the dark region in Figure 37).
Figure 37: Sample PIV image captured in cutwater region. Note uniform seeding and high-contrast.

Figure 38: PIV images showing synchronization consistency. Perpendicular impeller blades lie along the bottom and right edges of each image, demonstrating that the synchronization setup successfully captured images at the desired impeller positions.
Figure 39: Contours plots showing data obtained with PIV. Clockwise from top-left: relative speed with streamlines; principal Reynolds shear stress; viscous shear stress magnitude; vector quality. Operating point shown is 2800 RPM, 0.6 LPM (blade passage)
Figure 40: 2800 RPM, 3.0 LPM (blade passage)
Figure 41: 2800 RPM, 6.0 l PM (blade passage)
Figure 42: PRSS Distribution Summary in the blade passage
Figure 44: 2800 RPM, 3.0 LPM (back gap)
Figure 45: 2800 RPM, 6.0 LPM (back gap)
Figure 46: PRSS Distribution Summary in the back gap
Figure 47: Clockwise from top-left: speed with streamlines; principal Reynolds shear stress; viscous shear stress magnitude; vector quality. Operating point shown is 2800 RPM, 5.6 LPM (cutwater region)
Figure 48: 2000 RPM, 3.0 LPM (cutwater region)
Figure 49: 2800 RPM, 6.0 LPM (cutwater region)
Figure 50: PRSS Distribution Summary in the cutwater region
Hemolysis Testing

The measured rise in plasma free hemoglobin for each run of the FDA VAD using the polyurethane and Teflon shaft seals are shown in Figure 51 and Figure 52, respectively. The horizontal axes show test-time in minutes, and the vertical axes show the change in hemoglobin content in grams per liter. For each data set, the mean of the change in plasma free hemoglobin observed during the non-spinning seal tests was subtracted from that of the spinning tests to determine the hemolysis caused by the shaft seal interface alone. All data were normalized with respect to initial flow loop volumes. Standard error was prescribed to measurements for which two or more data points were collected. Lines were fit to the data in order to help distinguish between trends. VAD experiments are shown as circles and shaft seal experiments are shown as triangles. Note that the data of interest are the rates at which hemolysis occurs in each test rather than any particular measurement. The rates of hemolysis were taken as the slopes of the linear fits. The contribution of each seal to the total VAD hemolysis was taken as the rate of hemolysis due to the shaft seal divided by the rate of hemolysis due to the total VAD tests, as summarized in Figure 53. The rate of hemolysis due to the polyurethane seal was 39%-62% of that due to the total VAD, while the rate of hemolysis due to the Teflon seal was 7%-9% of the total VAD.
Figure 51: FDA VAD and shaft seal tester hemolysis data for Polyurethane shaft seal. Data were normalized with respect to initial blood volume. VAD tests are shown as circles and shaft seal data is shown as triangles. Standard error was prescribed to measurements for which two or more data points were collected.
Free plasma hemoglobin, Teflon seal

![Graph showing free plasma hemoglobin levels for Teflon shaft seal with data points and error bars.]  

6.0 LPM  
• 6.0 LPM  
• 0.6 LPM  
• Seal Damage  
- Linear (6.0 LPM)  
- Linear (0.6 LPM)  
- Linear (Seal Damage)

Time (minutes)  
0 30 60 90 120 150  
0 5 10 15 20 25

Figure 52: FDA VAD and shaft seal tester hemolysis data for Teflon shaft seal. Data were normalized with respect to initial blood volume. VAD tests are shown as circles and shaft seal data is shown as triangles. Standard error was prescribed to measurements for which two or more data points were collected.
Figure 53: Contribution of the shaft seal hemolysis to the total measured VAD hemolysis. For each flow rate, polyurethane seal data is shown on the left (blue) and Teflon seal data on the right (red). Data was not obtained for the Teflon seal at 3 LPM.
Further Observations

Post-test inspection revealed significant darkening of the dry-side of the polyurethane shaft seal, as shown in Figure 54. Another polyurethane seal emitted smoke when run dry, and the drive shaft became hot to the touch. Additionally, structures resembling burned were found at the shaft-seal interface after the polyurethane VAD tests. Similar formations were found after spinning shaft-seal tests, but were generally smaller than those formed in the VAD. Photographs of these structures found in the VAD and in the seal tester (for the polyurethane seal) are shown in Figure 55 and Figure 56, respectively. The shaft seal was superficially cleaned after each test, but the thrombus formations were tenacious and difficult to remove completely. Indications of excessive heating (seal discoloration and unusual blood formations) were absent in all tests using the Teflon shaft seal. A small thrombus-like formation was found after one test, but it did not resemble the “burned blood” seen in Polyurethane tests. A photograph of the Teflon seal after a VAD test is shown in Figure 57.

Figure 54: Discoloration of dry-side of polyurethane shaft seal observed after a day of testing.
Figure 55: Thrombus-like formation found at the shaft-seal interface after a VAD test (2800 RPM, 0.6 LPM)

Figure 56: Thrombus-like formation found at the shaft-seal interface after a shaft seal test (2800 RPM, 0.6 LPM)
Figure 57: Teflon shaft seal following a VAD hemolysis test. Note that the seal’s spring is still visible through the small thrombus-like formation.
Chapter 6: Discussion & Conclusions

Remarks

The pump demonstrated a range of flow fields containing interesting features including recirculation, shear layers, and turbulent wakes that are characteristic of rotary VADs. These features were operating point dependent, which demonstrated the pump’s utility for validation of CFD predictions. The pump design was robust and was constructed using transparent materials, allowing for visualization of blade passage, back gap, and cut-water over a range of flow rates.

Preliminary tests with the prototype FDA VAD showed that the contribution of the polyurethane shaft seal to the total VAD hemolysis was significant. The rate of hemolysis due to the Teflon seal, however, was in all cases below 10% of that measured in the VAD, which meets the design criteria for an acceptable shaft seal specified in Equation 48. Recalling the blood-pump design process regarding hemolysis, shown previously in Figure 2 and re-printed here for reference, the present research was executed in support of efforts to determine the capacity to which modern CFD is able to predict shear stress fields within blood pumps. Thus, the pump will serve its intended purpose for the FDA CPI project.

Figure 58: Typical blood pump design process with respect to blood damage, as shown in Figure 2.

The present research was successful in the execution of its predefined goals:

1. Design and manufacture a prototypical centrifugal blood pump throughout which the velocity and shear-stress distributions may be experimentally measured
   - A prototype centrifugal blood pump was designed and built in conjunction with the FDA working group. The pump was designed to be easily modeled for computational analysis. The hydraulic performance was measured to be
reasonably representative of a clinical blood pump. The pump was integrated into a robust and precisely controllable imaging platform for quantitative flow measurements.

2. Experimentally measure velocity and shear-stress fields within the pump
   - Particle image velocimetry was successfully used to obtain measurements of velocity and viscous and turbulent shear stress within the pump. Synchronization of blade position with image capture was achieved. The positioning hardware was demonstrated as a quick and precise method of selecting image locations. The flow fields depended strongly on the pump’s operating point.

3. Experimentally measure the hemolysis caused by the pump.
   - Hemolysis experiments were performed with the prototype VAD. Additional experiments were performed in an attempt to distinguish between hemolysis due to the impeller and that due to the shaft seal. The hemolysis produced by the VAD was appreciable. The contribution of the polyurethane shaft seal to VAD hemolysis was significant (above 10%), while the contribution of the Teflon seal to VAD hemolysis was small (below 10%).

**Pump Performance**

The measured pump performance indicates that the prototype FDA VAD delivers flow rates and pressures reasonably similar to those of a clinical VAD. The slight discrepancy between the data collected with water and with blood is most likely due to fact that blood is more dense and viscous than water. Bernoulli’s principle predicts that the pressure rise across a pump at any operating point is proportional to the fluid density. As expected, the maximum pressures measured and predicted with blood and water, respectively, were 250 mmHg and 235 mmHg, in a ratio of 1.06, which is close to the specific gravity of blood (1.05). Additionally, the collapse of the dimensional pump data to a single non-dimensional curve suggests that the non-dimensional analysis was executed successfully.

The drive motor and impeller shaft were initially joined by a flexible shaft coupler. This led to the discovery that the impeller generated a significant axial force toward the
upper housing, which hindered the ability to maintain the impeller’s axial position. While difficult to measure, the magnitude of this force is thought to be roughly equal to the pressure rise times the impeller area, which is about 16 pounds. In order to precisely maintain the impeller’s axial position, the flexible shaft coupler was replaced by a rigid coupler. This worked well initially, but required precise concentricity of all rotating parts and thus complicated pump assembly. While an axially stiff but radially flexible shaft coupler may appear as an attractive solution, maintaining the axial position within 1 mm would require a coupler with an axial stiffness of about 500 pounds per inch. This spring rate is typical of automotive suspension components and may be difficult to obtain in a miniature shaft coupler. Clearly, a more practical solution must be found. One attractive alternative could be to revisit the design requirement of a controllable impeller position. This requirement was necessary in the prototype VAD to determine the minimum spacing that would allow the capture of clear images in the back-gap region. Now that this has been determined to be 1 mm, however, the spacing may be fixed in future design iterations, alleviating the requirement of a rigid shaft coupler and easing the difficult of aligning the rotating hardware.

As noted in the results section, the pump’s stainless steel drive shaft was severely scarred at the seal interface after PIV experiments. No such wear was observed after blood damage experiments. The shaft scarring is therefore thought to be the result of the introduction of glass PIV particles into the shaft-seal interface and acting as an aggressive abrasive agent. This is reasonable considering that glass is about eight times harder than stainless steel. The proposed solution is to perform future PIV experiments with plastic seed particles, which are much softer than steel and thus should not damage the shaft.

**PIV**

The processed PIV data indicated high quality vectors throughout most of the flow. Localized maximum velocities, Reynolds stresses, and viscous stresses, however, were observed to correspond to regions of lower quality vectors, indicating that citation of peak values was not an accurate metric by which to compare data sets. Additionally, scaling the data to span the entire range of measured values returned poor resolution among contour plots. For these reasons, the data were presented with similar ranges which, in some cases, were far below maximum measured values. Although this method of presentation led to contour saturation, it was believed to constitute the most conservative means by which the flows’ behaviors could be compared. The pump produces principal Reynolds shear stresses over 250 Pa, which is predicted by some literature to cause hemolysis (Sutera and Mehrjardi...
As shown in Figure 39 - Figure 50, the magnitude and distribution of these stresses clearly depended on the pump's operating point, which was in line with the goals of the FDA CPI project.

The measured velocity fields and streamlines provided some insight regarding the pump's ideal operating point. Recirculation regions were present in the diffuser for both 0.6 and 6.0 LPM, while the mean flow appeared to remain attached to the diffuser at 3.0 LPM. This suggested that the pump's ideal operating point at 2800 RPM was around 3 LPM. Interestingly, consideration of the Reynolds stress fields and the mean velocity fields may lead to conflicting implications regarding the pump's potential to cause hemolysis. Based on the well-behaved flow in the diffuser at 3.0 LPM, it could follow that the flow at this operating point may be rather orderly throughout the pump and thus present the least potential for hemolysis. However, integration of the Reynolds stress histograms with respect to the number of data points in each range suggested that Reynolds stresses were weakest throughout the pump at 0.6 LPM and strongest at 6.0 LPM. This should not, however, be taken as a prediction of the relationship between pump operating point and the predicted hemolysis, since other factors (such as increased blood-cell residence time due to recirculation regions) must be taken into account. While interesting, the prediction of hemolysis based on Reynolds stress fields was beyond the scope of the present research, and was not attempted.

The important implication of the present PIV results was that the pump produced fields of Reynolds stress, velocity, and viscous shear stress that were strongly dependent, both quantitatively and qualitatively, on the pump's operating point, and may thus serve as a benchmarking tool against which future CFD predictions may be judged. In this respect, the prototype FDA VAD fulfilled the objectives of the CPI project.

**Hemolysis Testing**

The hemolysis data indicate that the VAD does cause appreciable hemolysis. The data suggest that the hemolysis caused by the isolated polyurethane shaft-seal interface is similar to that caused by the complete FDA VAD. It is thought that hemolysis due to this seal may be attributed to intense localized heat generation at the shaft-seal interface. The dark discoloration of the polyurethane shaft seal observed after testing suggests that the seal becomes sufficiently heated to burn, which supports this hypothesis. The polyurethane shaft seal was chosen for its FDA approval and durability, not its coefficient of friction; these seals were essentially precision-machined pencil erasers. It is therefore reasonable, in retrospect,
to expect that they would produce an appreciable amount of heat when gripping a steel shaft spinning close to 3000 RPM. The heat generated at the shaft seal, then, could be reduced by selecting a seal material with a lower coefficient of friction than polyurethane. In support of this hypothesis, the contribution of the Teflon shaft seal to the total VAD hemolysis was indeed much less than that of the polyurethane seal. This indicates that the majority of the hemolysis in VAD experiments using the Teflon seal was due to the components that may be modeled using CFD. Thus, the VAD is suitable for comparison to CFD predictions in support of the FDA CPI objectives.
Works Cited


Appendices

Appendix A: FDA VAD Design Decision Process

Several designs were considered for various parts of the pump. Designs that were incorporated into the final pump are shown in green.

Suspension

- **Magnetic coupling / jewel bearings**
  - Pros:
    - No shaft seal
    - Minimal heat generation
    - Small contact area minimizes hemolysis
    - Easier for CFD models
  - Cons:
    - Difficult to assemble
    - Jewels obstruct inlet
    - Unsure about torsional stiffness of magnetic coupling

- **Magnetic coupling / internal ball bearings**
  - Pros:
    - Proven design (Biomedicus)
    - Smaller shaft seal
    - Easy to assemble
    - Design may be borrowed from Biomedicus pump
  - Cons:
    - Unsure about torsional stiffness of magnetic coupling
    - Internal shaft and seal

- **Air Supply Seal**
  - Pros:
    - Reduces shear at seal
  - Cons:
    - Overly complicated

- **Ferrofluid Seal**
  - Pros:
    - Reduces shear at seal
  - Cons:
    - Overly complicated
    - Ferrofluid leaks into working fluid
    - Requires air supply

- **Saline in Seal Gap**
  - Pros:
    - Helps seal operation
• Cons:
  • Overly complicated
  • Seal leaks

• Through-wall shaft / ball bearings
  • Pros:
    • Simple design
    • Easy to manufacture
    • Rigid in torsion / precise knowledge of angular position at all speeds
    • Easy to assemble
    • Small shaft present in mag-coupled design anyway
    • Small shaft should cause minimal hemolysis
    • No internal bearings
  • Cons:
    • Thru-wall shaft difficult for CFD modeling
    • Heat generation / hemolysis at seal

Driver
  • Med-Tronic Driver
    • Pros:
      • In RIT inventory
      • Compatible with magnetically coupled pump
    • Cons:
      • Complicates optical setup
      • Limited shaft speed
      • Constrains pump dimensions

• Electric motor rig
  • Pros:
    • Does not constrain optical setup
    • Does not constrain pump design
    • Compatible with both shaft driven and magnetically coupled pumps
  • Cons:
    • Potentially more construction

Volute
  • Logarithmic volute profile
    • Pros:
      • Representative of most centrifugal blood pumps
      • Better pressure recovery than circular profile
    • Cons:
      • Complicate manufacturing (requires CNC)
      • Complicates computational modeling
      • Maximizing pressure recovery is not important

  • Circular volute profile
    • Pros:
      • Ease of machinability
- Ease of computational modeling
- Closely approximates log volute based on pump design analysis
  - **Cons:**
    - Less efficient than logarithmic profile
    - Not as representative of many centrifugal blood pumps

**Impeller Blood Washout**

- **Washout holes through impeller**
  - **Pros:**
    - Reduce thrombus formation
  - **Cons:**
    - May be unnecessary due to relatively short run times
    - Complicate manufacturing
    - Thrombus formation may be desirable for CFD validation

- **Spokes**
  - **Pros:**
    - Reduced thrombus formation
  - **Cons:**
    - Complicates fabrication
    - Not representative of many centrifugal blood pumps

- **Blades on underside of impeller or hub**
  - **Pros:**
    - May reduce thrombus formation
  - **Cons:**
    - Design not representative of many centrifugal blood pumps

- **No washout, impeller seals to shaft on ID**
  - **Pros:**
    - Simple to manufacture and model
    - Short run times may not lead to thrombus formation anyway
    - Thrombus formation could be used for CFD validation
  - **Cons:**
    - Thrombus formation

**Impeller Hub and Shroud**

- **Hub on bottom/ shroud on top**
  - **Pros:**
    - High strength
    - Allows for drive magnets
  - **Cons:**
    - Optically occludes blade passage (unless shroud is clear)
    - Complicates manufacturing
    - May reduce turbulence intensity

- **Shroudless and hubless impeller**
  - **Pros:**
• Reduce thrombus formation
  • Compatible with aluminum shaft driven impeller
  ▪ Cons:
    ▪ Leaves minimal space for drive magnets (if mag-drive)
    ▪ Reduces blade strength

• Only bottom hub
  ▪ Pros:
    ▪ Increases blade strength
    ▪ Representative of many centrifugal blood pumps
    ▪ Leaves room for mag-drive magnets
  ▪ Cons:
    ▪ Thrombus formation underneath hub
    ▪ Optically occludes space behind hub (unless hub is clear)

ImPELLer SSHAPE

• Curved / complex blade shape
  ▪ Pros:
    ▪ Representative of real centrifugal blood pumps
    ▪ Better pump performance
  ▪ Cons:
    ▪ Complicates computational modeling

• Straight blades
  ▪ Pros:
    ▪ Ease of manufacturing
    ▪ Ease of computational modeling
    ▪ Cause moderate hemolysis in literature
  ▪ Cons:
    ▪ Not representative of most centrifugal blood pumps
    ▪ Risk of cavitation

ImPELLer Material

• Stainless Steel
  ▪ Pros:
    ▪ Relatively easy to machine (CNC)
    ▪ High strength (can make hubless impeller)
    ▪ Biocompatible
  ▪ Cons:
    ▪ Opaque
    ▪ Optically occludes part of pump

• Aluminum
  ▪ Pros:
    ▪ Easy to machine
    ▪ High strength (can make hubless impeller)
  ▪ Cons:
    ▪ Opaque
    ▪ Optically occludes part of pump
• Biocompatibility concerns

- Acrylic
  - Pros:
    - Clear
    - Can image entire pump with single laser setup
  - Cons:
    - Brittle (needs shrouded impeller)
    - More difficult to machine

Inlet Design

- Curved Tubing
  - Pros:
    - Representative of real inlet conditions (cannula)
    - Does not interfere optical setup as severely
  - Cons:
    - Could complicate inlet boundary condition specification
    - Was unable to bend stainless steel tubing without kinking

- Straight Tubing
  - Pros:
    - Ease of manufacturing
    - Ease of computational modeling
  - Cons:
    - Could obstruct optical setup unless kept short
Appendix B: Engineering Drawings

(Continued on following pages)
Slip Fit to 1/8" Shaft
Ø0.155 ±0.001
Ø0.354 ±0.001
Ø0.315 ±0.001
0.000
0.630
0.000

Ø0.089 ±0.001
Bottoming 4-40 Tap

Stainless
Polished
Motor mount top

Aluminum
Appendix C: Bills of Materials
(Continued on following pages)
<table>
<thead>
<tr>
<th>Product Description</th>
<th>Part Number</th>
<th>Cost (each)</th>
<th>Qty</th>
<th>Total</th>
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<td>Stainless Steel Tubing (10mm x 14mm x 8&quot;)</td>
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<td>Rigid shaft coupling</td>
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<tr>
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<tr>
<td>Pump Housing (lower)</td>
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<td>Unifab, Inc</td>
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<td>Impeller</td>
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<td>Hardware</td>
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Total (FDA VAD) $3,131.93
Grand Total $4,214.36
### Motor Controller Assembly

<table>
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<tr>
<th>Product Description</th>
<th>Vendor</th>
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<th>Cost (each)</th>
<th>Qty</th>
<th>Total</th>
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**Total** $451.45
Appendix D: PIV Statistical Analysis Code (MATLAB)

Ensemble Statistics Code

function statistics_speed_pump(drive,experiment,run,start,stop)
% STATISTICS_SPEED Statistical Vector File Analyzer
% statistics_speed_pump(drive,experiment,run,start,stop) =
% performs statistical analysis of a batch of PIV vector files.
% INPUT
% drive: Letter of drive on which PIV data is stored
% experiment = name of experiment folder under which PIV data is stored
% run = name of PIV run under which PIV data is stored
% start: First image to be processed
% stop: Last image to be processed
% OUTPUT
% x = axial coordinate (pixels)
% y = radial coordinate (pixels)
% umean = mean horizontal component of velocity (pixels / frame)
% vmean = mean vertical component of velocity (pixels / frame)
% ustd = standard deviation of horizontal component of velocity (pixels / frame)
% vstd = standard deviation of vertical component of velocity (pixels / frame)
% tij = shear rate (1 / frames)
% ruu = horizontal component of Reynolds normal stress (pixels² / frame)
% rvv = vertical component of Reynolds normal stress (pixels² / frame)
% ruv = Reynolds shear stress (pixels² / frame)
% vort = vorticity (1 / frames)
% count = number of valid vectors
% EXAMPLE
% Experimental Parameters:
% First vector file:
% C:\Experiments\fdavad\2800RPM\posl_600mlpm_vector\posl_600mlpm_-B-000000.plt
% Final vector file:
% C:\Experiments\fdavad\2800RPM\posl_600mlpm_vector\posl_600mlpm_-B-000999.plt
% Function Inputs:
% drive = 'C';
% experiment = '2800RPM';
run = 'posl_600mlpm_';
start = 0;
stop = 999;

See also std, mean, curl, gradient, reshape

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Department of Mechanical Engineering
matthew.giarra@gmail.com
8 May 2009

Start timer
filestart = cputime;

% Strings identifying data paths
sequence = [drive '\Experiments\fdavad\' experiment '\' run '\vector\'];
series = [run '-B-'];

writeto = sequence;
% Check existence of directory; create directory if it does not exist
if exist(writeto, 'dir') == 0;
    mkdir(writeto);
end

% Print status
fprintf(1,'\n\nProcessing series %s ... \n \n', series);

% File Information
extension = '.avg';
target = [series num2str(stop - start) extension];
fid = fopen([writeto target], 'w');

% Load raw PIV data
fprintf(1,'\nLoading vector files... \n \n');
[raw, depth] = fileloader(sequence, series, start, stop);

% Determine matrix size
m = size(raw(:,1,1));
%
% Populate position vectors
x = raw(:,1,1);
y = raw(:,2,1);
%
% Name other columns of raw data
% Velocities
u = raw(:,3,:);
v = raw(:,4,:);
%
% Vector quality
info = raw(:,7,:);
% Create zero matrices for averaged values
% Velocities
umean = zeros(m,1);
vmean = zeros(m,1);
% Standard deviations
ustd = zeros(m,1);
vstd = zeros(m,1);
% Speed
speed = zeros(m,1);
% Reynolds Stresses
ruu = zeros(m,1);
rvv = zeros(m,1);
ruv = zeros(m,1);
% Number of valid vectors
count = zeros(m,1);
usum = zeros(m,1);
% Print status
fprintf(l,'Performing statistics... \n \n');
% Perform statistics
for i = 1:m;
  for k = 1:depth
    if info(i,k) > 0;
      u(i,k) = 0;
      v(i,k) = 0;
    end
  end
  count(i) = length(nonzeros(u(i, :)));
  if count(i) < 2;
    umean(i) = 0;
    vmean(i) = 0;
    ustd(i) = 0;
    vstd(i) = 0;
  else
    % Determine mean velocities
    umean(i) = mean(nonzeros(u(i,:)));
    vmean(i) = mean(nonzeros(v(i,:)));
    % Determine standard deviations of velocities
    ustd(i) = std(nonzeros(u(i,:)));
    vstd(i) = std(nonzeros(v(i,:)));
    % Determine Reynolds normal stresses
    ruu(i) = -1 * (ustd(i))^2;
    rvv(i) = -1 * (vstd(i))^2;
    % Determine mean speeds
    speed(i) = (umean(i)^2 + vmean(i)^2 )^0.5;
  end
end
% Calculate Reynolds Shear Stresses
for k = 1:depth;
    if u(i,k) == 0;
        % Calculate Reynolds Stresses
        usum(i) = usum(i) + ((u(i,k) - umean(i)) * (v(i,k) - vmean(i)));
    end
end
end
if count(i) > 1
    ruv(i) = -1 * usum(i)/count(i);
else
    ruv(i)=0;
end

% Create 2-D velocity maps to be used by the "Curl" function to calculate vorticity
width = max(x)/16;
height = max(y)/8;
umap = reshape(umean,width,height);
umap = umap';
vmap = reshape(vmean,width,height);
vmap = vmap';

% Calculate vorticity
vortmap = curl(umap,vmap);

% Calculate velocity field gradients
[dudx dudy] = gradient(umap);
dudy = 8 * dudy;
dvdx = 16 * dvdx;

% Calculate shear rate
tij = (dudy + dvdx);

% Put vorticity matrix back into vector form for compatibility with Tecplot
vort = reshape(vortmap',m,l);
tij = reshape(tij',m,l);

% Write header to be read by Tecplot
fprintf(fid, 'TITLE="%s%s"
',sequence,series);
fprintf(fid, 'VARIABLES="X","Y","Umean","Vmean","Speed","Ustd","Vstd","Shear Rate","Ruu","Ryy","Ruv","Vort","Count"
');
fprintf(fid, 'ZONE T="Velocity field", I=79, J=127\n');

% Create matrix containing all data to be written to file
data = [x,y,umean,vmean,speed,ustd,vstd,tij,ruu,rvv,ruv,vort,count];

% Print status
fprintf(1, 'Writing data file...

');

% Write data to target text file
for i = 1:size(data,1);
    for j = 1:size(data,2);
        fprintf(fid, '%6.5f\t', data(i,j));
    end;
    fprintf(fid, '
');
end

% Close data file
fclose('all');

% Calculate performance
fileend = cputime;
elapsed = fileend - filestart;
speed = depth / elapsed;

% Report performance
fprintf(1, '\n Processing complete! \nFile saved to %s%s 
', writeto, target);
fprintf(1, 'Elapsed time: %6.2f seconds \n', elapsed);
fprintf(1, 'Average speed: %6.2f files per second \n', speed);

Supplementary code: Vector File Loader

function [data depth] = fileloader(sequence,series,start,stop)
    % FILELOADER Vector File Loader.
    % fileloader(sequence,series,start,stop) = [data depth]
    % Loads a series of PIV vector files for statistical analysis by
    % the function statistics_speed
    % INPUT
    % sequence = directory in which vector files are located.
    % series = prefix of vector series files.
    % start = number designating initial vector file.
    % stop = number designating final vector file.
    % OUTPUT
    % data = M-by-N-by-K array of vector data
    % M = total number of PIV interrogation regions in each vector file
    % N = number of variables output by PIV processing code
    % K = number of vector files to be processed
    % depth = number of vectors files to be processed
    % See also dlread
    % Matthew M. Giarra
    % Rochester Institute of Technology
    % Department of Mechanical Engineering
    % matthew.giarra@gmail.com
    % 8 May 2009
% Depth of 3-D matrix containing all data
depth = stop - start + 1;

% Create empty matrix to contain file numbers
images = zeros(depth,1);
images(1) = start;

% Populate matrix with file numbers
for i=2:depth,
    images(i) = images(i - 1) + 1;
end

% Define filename leaders and extensions
leader5 = '00000';
leader4 = '0000';
leader3 = '000';
leader2 = '00';
extension = '.plt';

% Create and populate matrix containing series of file names
for i = 1: depth
    if images(i) < 10
        filepath(i,:) = {'sequence seriesleader5 int2str(images(i)) extension'};
    elseif images(i) >= 10 && images(i) < 100
        filepath(i,:) = {'sequence seriesleader4 int2str(images(i)) extension'};
    elseif images(i) >= 100 && images(i) < 1000
        filepath(i,:) = {'sequence seriesleader3 int2str(images(i)) extension'};
    else
        filepath(i,:) = {'sequence seriesleader2 int2str(images(i)) extension'};
    end
end

% Read first vector file
raw = dlmread(char(filepath(1,:)),' ',3,0);

% Determine dimensions of vector files
[rawr rawc] = size(raw);

% Allocate memory for 3-D matrix to accommodate vector series
data = zeros(rawr, rawc, depth);

% Read vector series series
for i = 1:depth;
    data(:,:,i) = dlmread(char(filepath(i,:)),' ',3,0);
end;

PIV Calibration Code
function pumpcalibrator(experiment, run, num, drive, spacecal, dvolute, xshiftmm, yshiftmm, dt, omega, rho, visccp)

% PUMPCALIBRATOR PIV Data Calibrator

pumpcalibrator(experiment, run, num, drive, spacecal, dvolute, xshiftmm, yshiftmm, dt, omega, rho, visccp) =

% [x, y, uvector, vvector, speedvector, urelvector, vrelvector, speedrelvector, uthetavect, ustd, vstd, ss, prss, prns, vort, count]
% Converts rotational PIV data to SI units.

% INPUT
% drive: Letter of drive on which PIV data is stored
% experiment = name of experiment folder under which PIV data is stored
% num = number of images used to produce ensemble averaged vector file to be calibrated
% spacecal = Spatial calibration (microns/pixel);
% dt = Laser Pulse Spacing (microseconds);
% omega = Shaft speed (RPM);
% rho = Fluid density (kg/m^3);
% visc = Dynamic viscosity (cP);
% xshift = Horizontal offset of impeller with respect to image origin (mm);
% yshift = Vertical offset of impeller with respect to image origin (mm);
% dvolute = Volute diameter (mm);

% OUTPUT
% x = Horizontal position (mm);
% y = Vertical position (mm);
% uvector = Horizontal component of velocity (m/s);
% vvector = Vertical component of velocity (m/s);
% speedvector = Absolute speed (m/s);
% urelvector = Horizontal component of velocity relative to impeller's rotating reference frame (m/s);
% vrelvector = Vertical component of velocity relative to impeller's rotating reference frame (m/s);
% speedrelvector = Speed relative to impeller's rotating reference frame (m/s);
% uthetavect = Tangential velocity (m/s);
% ustd = Standard deviation of horizontal component of velocity (m/s);
% vstd = Standard deviation of vertical component of velocity (m/s);
% ss = Viscous shear stress (Pascals);
% prss = Principal Reynolds shear stress (Pascals);
% prns = Principal Reynolds normal stress (Pascals);
% vort = Vorticity (1/seconds);
% count = Number of valid vectors

EXAMPLE

Experimental Parameters:

Vector file path: C:\Experiments7\fdavad\2800RPM\
Spatial calibration = 29.35 microns/pixel;
Laser Pulse Spacing = 56 microseconds
Shaft speed = 2800 RPM;
Fluid density = 1862 kg/m$^3$;
Dynamic viscosity = 4.7 cp
Horizontal offset of impeller with respect to
image origin = -1.97 mm;
Vertical offset of impeller with respect to
image origin = -1.45 mm;
Volute diameter = 60.0 mm;

Function Inputs:
  drive = 'C';
  experiment = '2800RPM';
  run = 'pos1_6000mlpm_2';
  num = 999;
  spacecal = 29.35;
  dt = 56;
  omega = 2800;
  rho = 1862;
  visc = 4.7;
  xshift = -1.97;
  yshift = -1.45;
  dvolute = 60;

See also dlmread, reshape, statistics_speed_pump

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Strings identifying data paths
sequence = [drive '\Experiments\fdavad\' experiment '\run \vector\' run '-B-');
series = [run '-B-');
extension = '.avg';

File Information
target = [sequence num2str(num) extension];
writetarget = [sequence '_cal_' num2str(num) extension];
raw = dlmread(target, '\t', 3, 0);

Convert horizontal shift to meters
xshift = xshiftmm / 1000;

Convert vertical shift to meters
yshift = yshiftmm / 1000;

Convert volute diameter in millimeters to volute radius in meters
rvolute = (dvolute / 2) / 1000;
% Convert viscosity from centipoise to mPa*s
visc = visccp / 1000;

% Specify camera CCD resolution
xpixels = 1280;
ypixels = 1024;

% Determine number of interrogation regions assuming 16 and 8 pixels per
% region in horizontal and vertical directions, respectively. This should
% be changed if PIV processing resolution changes.
xcells = (xpixels / 16) - 1;
ycells = (ypixels / 8) - 1;

% Extract relevant data from .avg file
x = raw(:, 1);
y = raw(:, 2);
u = raw(:, 3);
v = raw(:, 4);
speed = raw(:, 5);
ustd = raw(:, 6);
vstd = raw(:, 7);
tij = raw(:, 8);
ruu = rho * raw(:, 9);
rvv = rho * raw(:, 10);
ruv = rho * raw(:, 11);
vort = raw(:, 12);
count = raw(:, 13);

% Determine viscous shear stress (Pa)
ss = abs(tij * visc / (dt / 10^6));

% Determine Reynolds stresses relative to PIV coordinate system (Pa)
ruucal = ruu * (spacecal / dt)^2;
rvvcal = rvv * (spacecal / dt)^2;
ruvcal = ruv * (spacecal / dt)^2;

% Determine principal Reynolds stresses (Pa)
prss = ((ruucal - rvvcal)/2) + ruvcal;(0.5);
prns = (ruucal + rvvcal)/2 + (0.5 * (ruucal - ruvcal)*2 + ruvcal^2)(0.5);

% Apply spatial calibration to convert cartesian positions from pixels to
% meters
xcal = x * spacecal / 10^6;
ycal = y * spacecal / 10^6;

% Apply velocity calibration to convert cartesian velocities to m/s
ucal = u * spacecal / dt;
vcal = v * spacecal / dt;
umap = (reshape(ucal, xcells, ycells))

% Create 2-D cartesian coordinate maps (meters)
ycal = (reshape(ycal, xcells, ycells));
% Perform cartesian to cylindrical coordinate transform
[theta,r] = cart2pol(xmap - xshift, ymap - yshift);
theta = abs(theta);

% Calculate radial velocity
ur = vmap.*sin(theta) + umap.*cos(theta);
urvreft = reshape(ur,xcells*ycells,1);
figure
% contourf(xmap,ymap,urmap);
% title 'Contours of radial velocity (m/s)'

% Calculate circumferential velocity
utheta = vmap.*cos(theta) - umap.*sin(theta);

% Allocate memory for relative tangential velocity
uthetarel = zeros(length(utheta),1);

% Determine relative tangential velocity
for i = 1:ycells;
    for j = 1:xcells;
        if r(i,j) < rvolute;
            uthetarel(i,j) = utheta(i,j) - omega * 2*pi / 60 * r(i,j);
        else
            uthetarel(i,j) = utheta(i,j);
        end
    end
end

% Transform velocity components back into cartesian coordinates
u = umap;
urvreft = reshape(u',xcells*ycells,1);
vrel = reshape(vmap',xcells*ycells,1);
speedvreft = reshape(speed',xcells*ycells,1);
urel = (ur - uthetarel.*tan(theta))./(sin(theta).*tan(theta) + cos(theta));
vrelvrel = reshape(vrel',xcells*ycells,1);
urelurel = reshape(urel',xcells*ycells,1);
urvrelt = reshape(urrelvrel,urelurel,vrelvrel,urelurel,uthetarel,uthetarel,xcells*ycells,1);
speedrel = reshape(speedrel',xcells*ycells,1);

% Convert spatial coordinates to millimeters
x = xcal * 1000;
y = ycal * 1000;

% Write header to be read by Tecplot
fid = fopen(writetarget, 'w');
fprintf(fid, 'TITLE="%s"
', writetarget);
fprintf(fid, 'VARIABLES="X (mm)", "Y (mm)", "U (m/s)", "V (m/s)", "Speed (m/s)", "Urel (m/s)", "Vrel (m/s)", "Speed_rel (m/s)", "Ur (m/s)"");
(m/s), "Ustd (m/s)", "Vstd (m/s)", "SS (Pa)", "PRSS (Pa)", "Vort (1/s)", "Count"
');
fprintf(fid, 'ZONE T="Velocity field", I=79, J=127\n');

% Create matrix containing all data to be written to file
data = [x, y, uvect, vvect, speedvect, vrelvect, speedrelvect, urvect, uthetavec, ustd, vstd, ss, prss, prns, vort, count];

% Print status
fprintf(l, 'Writing data file... \n \n');

% Write data file to target text file
for i = 1:size(data,1)
    for j = 1:size(data,2)
        fprintf(fid, '\%6.5f\t', data(i,j));
    end
    fprintf(fid, '\n');
end

% Close data file
fclose all;

% Print status
fprintf(l, 'Done! \n');