I, Suryanarayana Pappu, hereby submit this original work as part of the requirements for the degree of Master of Science in Mechanical Engineering.

It is entitled:
A Parametric Study to Quantify the Pressure Drop of Pulsating Flow through Blockages

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A Parametric Study to Quantify the Pressure Drop of Pulsating Flow through Blockages

A thesis submitted to the
Graduate School
of the University of Cincinnati
in partial fulfillment of the
requirements for the degree of

Master of Science
in the Department of Mechanical Engineering
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by

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Abstract

The current set up is an improvement of a previous set up designed and fabricated to study pulsating flow through blockages[1]. The ultimate objective of such an experiment is to non-invasively be able to detect blockages in a pipe line with such theory and results then extended to detecting blockages in arteries. This facility additionally has the capability of pulsing both upstream as well as downstream of the flow, that is, in the direction as well as the opposite to the direction of the flow to be able to study the effect of pulsing flow better and reach the ultimate goal of a near perfect facility in the future with improvements.

The facility has been designed such that pressure measurements can be made at different axial locations along the pipeline. Parameters such as pressure head, frequency of pulsation blockage size and location of pulsation are varied to study the various cases. In addition to this, the duty cycle of the pulse can be changed depending on the systole and diastole ratios. Different wheels have been designed to change the pattern of pulsing for a set specific frequency.

The results are looked at in the light of these parameter changes. The fast fourier transforms and pressure time traces are studied. The FFTs show typical fundamental tones at the pulsing frequencies with subsequent harmonics and a gradual roll off after about 20-30Hz. The pressure time traces give us a picture of the pressure cycle during the systole and diastole periods and the fluctuation during those periods. From these basic plots the fundamental pressures are obtained and pressure drops and percentage pressure drops across various blockages and for various pressure heads. As a general observation, the pressure drops are higher with higher pressure heads and greater extent of blockage. With a higher frequency of pulsation the percentage
pressure drop reduces and with a more accurate systole diastole ratio the percentage drop is even lower. Upstream pulsing shows a pressure drop and downstream pulsing a pressure rise. A study of the flow rates is also done and it is seen that the pressure drops and flow rates change proportionally for the different blockage cases. The design of the new wheels based on Systole/Diastole ratios from literature allowed for the simulation of a more accurate duty cycle and a pressure curve that matches the aortic curve more accurately.

Finally flow visualization is also carried out to study the flow. Vortices are seen downstream of the blockage. In the 96% case there is instant mixing once the dye reaches the downstream side. Finally a conclusion is drawn and some suggestions for future work have been given.
Acknowledgements

I firstly thank my advisor Dr Ephraim Gutmark for allowing me to conduct research on the topic pressure drop of pulsating flow through blockages. Without his ideas and guidance this work could never have been achieved.

Next I give thanks to Dr Jeff Kastner who has been highly instrumental in helping me through this project from the beginning right to the very end. Words cannot express how thankful I am as he put in countless hours helping me through the stages of design, fabrication, data acquisition, the analysis of the results and finally the presentation and the thesis document. I could not have achieved this level of success without Dr Jeff Kastner’s guidance and motivation.

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1 Introduction

1.1 Coronary Artery Disease [1]
Almost 7 percent of all Americans have heart disease and more than half the population suffering from heart disease are below the age 65.[2] Coronary Artery Disease (CAD) is one type of heart disease which is characterized by the narrowing of the arteries which results in the reduction of blood and oxygen to the heart. When the artery narrows down to a critical level the supply of oxygen and blood becomes so low that there is chest pain and the occurrence of a heart attack. CAD occurs when the arteries become blocked due to plaque deposition which is due to the build-up of cholesterol.

The first manifestation of CAD is either acute angina or stable angina. There are diagnostic challenges which begin from the non-acute phase. It starts with risk assessment based on symptoms, medical history and various risk factors which typically include patient specific habits and information.
A few common diagnostic methods are presented in the Table 1-1.

<table>
<thead>
<tr>
<th>Diagnostic Method</th>
<th>Sensitivity</th>
<th>Specificity</th>
<th>Cost</th>
</tr>
</thead>
<tbody>
<tr>
<td>Coronary angiography</td>
<td>Golden standard</td>
<td></td>
<td>$1,047.19</td>
</tr>
<tr>
<td>CT-Coronary angiography</td>
<td>97%</td>
<td>87%</td>
<td>$694.32</td>
</tr>
<tr>
<td>Echo exercise test</td>
<td>86%</td>
<td>81%</td>
<td>$247.05</td>
</tr>
<tr>
<td>Myocardial Perfusion Imaging (Exercise SPECT)</td>
<td>87%</td>
<td>73%</td>
<td>$449.48</td>
</tr>
<tr>
<td>ECG stress test</td>
<td>68%</td>
<td>77%</td>
<td>$94.39</td>
</tr>
</tbody>
</table>

Table 1-1 Various Diagnostic Methods[3]

Some are invasive like Coronary angiography which is the ‘Golden standard’ [3] for CAD diagnosis and is used as a reference test for validation of other test methods. But on the downside it is costly, invasive and exposes the patient to radiation. Coronary Tomography (CT) is a test which is expected to play a major role in the future because it is a non-invasive technique with high accuracy and sensitivity. However, during CT the patient is exposed to radiation, and it is still considered high cost. ECG is the easiest and most affordable, but its low accuracy and sensitivity means that it doesn’t necessarily return positive results even when the patient is in actual fact affected by CAD.

Even though there are a wide range of tests, there still remain diagnostic challenges.

1. A study done by the American College of Cardiology showed that 41% of patients without any history of CAD that were referred for coronary angiography were actually affected by obstructive CAD, obstructive CAD being defined as a 50% reduction in diameter which
translates to 75% area blockage. Therefore better strategies for risk stratification are required in order to reduce the negative coronary angiographies. This requires better application of non-invasive techniques.[3]

2. Asymptomatic CAD- Statistics show that 15% cases of Myocardial Infarction are fatal in the US and 39% worldwide. CAD needs to be diagnosed early before it can progress to a critical state. But only 18-43% of coronary attacks are preceded by angina and statistics suggest that 50% men and 64% women who die do not show any symptoms of it before death. Therefore it is not symptoms alone that CAD identification can be based upon, instead non-invasive approaches need to be used to test patients with high risk.

3. Affordability- Low cost and fast to use diagnostic tests need to be available in developing countries especially where CAD prevalence is high.

Basically low cost and low risk tests with higher accuracy than ECG test will improve the risk stratification before invasive tests such as coronary angiography can actually be used which will then reduce the number of negative results. With accuracy, cost effectiveness, and the ability to detect CAD at an early stage being the focal point of research, non-invasive techniques have become important and amongst them identifying acoustic signatures caused due to turbulence produced by the occlusions is one method that is very promising.[4]

1.2 Functioning of the Heart

Before getting into how the flow in arteries is disturbed due to blockages, a basic knowledge about the functioning of the heart, its basic cycle, and the stages involved is necessary. The cardiac cycle can be divided into two phases [5]: systole and diastole. When the heart is undergoing diastole the ventricles are relaxed and blood flows passively into the left and right ventricles from the left and right atria, respectively.
Once the ventricles fill up, they begin to contract in order to pump blood into the aorta and pulmonary artery. This marks the beginning of the systole phase. The atrioventricular valves are closed during this phase so that the blood from the ventricles doesn’t re-enter the atria and it is during this phase that the aortic valve and the pulmonic valve open allowing blood into them from the left and right ventricles, respectively. During the diastole phase the aortic and pulmonic valves are closed. Figure 1-1 shows the directions of blood flow during the systole and diastole phases. The aortic pressure is measured by inserting a catheter into the aorta from a peripheral artery.
Figure 1-2 is the cardiac cycle diagram which shows the various pressure curves during the systole and diastole phases. AP implies aortic pressure, LVP implies left ventricular pressure, LAP implies left aortic pressure. Also shown is the left ventricular volume changing during the two phases. All this is during a single cycle of cardiac relaxation and contraction. It is clearly seen that the aortic pressure rises during the systolic phase and drops gradually in the diastolic phase which is in tune with how the heart functions as regards to contraction and relaxation. It must be emphasized that the systole and diastole explained above are physiological. Systole and Diastole are also demarcated based on heart sounds rather than physiological events, systole being the period between the first (closing of the
bicuspud and tricuspid valves) and the 2\textsuperscript{nd} heart sounds (closing of the aortic valve). The remainder automatically is the cardiologic diastole. Figure 1-3 shows the systole and diastole length varying based on heart rate.

![Figure 1-3: Systole Diastole durations][4]

Now that the basic functioning of the heart is understood it is paramount to understand why and when heart attacks occur and their relation to the duration of systole and diastole phases and heart rate. In a study [6] the systole-diastole duration and ratio was studied as a function of heart rate in order to understand the functionality of the heart. Typically with the human heart the duration of systole is shorter than diastole and hence the systole/diastole ratio (S/D) is less than 1. It is observed in fact that patients with heart failure, a “prolongation of the left ventricular systole and shortening of the diastole” [7]. This mismatch is heightened during strenuous exercise and it is observed that heart attacks do occur quite often at higher heart rates. This study[6] used an acceleration sensor housed in a case. It was positioned near the mid sternal precordial region and was fastened. The experiment was conducted in a certain age group, both healthy as well as diseased patients, one of the diseases being CAD. A plot of how the S/D ratio changes based on heart rate is presented for patients with various sorts of heart conditions in Figure 1-4.
Figure 1-4: S/D ratios for various heart conditions[6]

The CHD (CAD) plot on the top right corner is the one that most pertains to the current work. With this data one can better understand how the systole and diastole are affected by heart rate. Figure 1-4 shows the systole diastole ratio of patients with various conditions such as HYP: systemic hypertension, CHD (CAD): coronary heart disease, VALV: valvular heart disease, DC: dilated cardiomyopathy and COPD: chronic obstructive pulmonary disease. In all these conditions the trend is similar in terms of S/D ratio increasing with increasing heart rate and in most cases it even goes above 1, which implies the aortic valve is open longer than closed. The top left picture NL is for normal control subjects.
1.3 Detection
As mentioned previously of all the non-invasive techniques, identifying blockages based on acoustic signatures produced due to turbulence near a blockage is most promising. This method is a signal to noise ratio study where the signal needs to be amplified in comparison with the noise that is produced due to the opening and closing of the valves. Typically sounds in the coronary artery are too soft to detect using a normal stethoscope. Microphones need to be very sensitive and ideally need to detect weak signals at the body surface. Cardiac microphones measure the pressure waves due to the chest acceleration or displacement. In addition to the chest wall, tissues, heart wall and body fat, the mechanical loading of the transducer from the chest wall distorts the murmurs at the chest surface [4]. At higher frequencies microphones become more sensitive due to loading effects. Increase in the mass of the sensor increases the attenuation and because of this loading effect recording noises above 500Hz is difficult.

1.4 Studies and Modeling
Models were tried and developed for the likely acoustic signal expected to be generated by constricted blood vessels. Basically when blood flows through the stenosis, normal forces to the artery wall are produced, and corresponding to this low frequency vibrations are transmitted through the body tissue and to the external chest microphone. Studies by Borisyuk et al [8] suggest that depending upon the extent of constriction the fluctuation pattern of the blood flow changes. The greater the severity of the stenosis, greater is the sound generated due to increased turbulence. Spectral characteristics showed that the resonant frequencies were different in stenosed arteries as compared to normal arteries. In both there are low frequency components but in stenosed arteries there is a distinct higher frequency component which when the degree of stenosis increases shifts higher, with the low frequency component shifting to a lower frequency.
At 85% blockage, the shift was even more marked. An area obstruction of 85% causes a significant pressure drop that affects arterial blood flow [9].

It has been shown [10] that there is a relation between the diameter of the artery, velocity of flow, local turbulence and wall pressure fluctuations. It has also been identified that sounds associated with the turbulence in the artery are broadband in nature and typically roll off after a particular identifiable ‘break frequency’.

![Figure 1-5: Roll off after ‘Break Frequency’][10]
Figure 1-5 shows the sound spectrum of a typical post stenotic arterial bruit. The intensity of sound increased to a discrete maximum after which it fell off as frequency increased further. The frequency at which the peak amplitude occurs is called the break frequency $f_0$. 500Hz in this case. Each point in the plot represents the mean amplitude of a 20Hz segment from a digital Fast Fourier Transform (FFT).

In analyzing these sounds, since eliminating the heart valve sounds is paramount it is essential to isolate that segment without the ‘lub’ ‘dub’ sounds. Figure 1-6 shows the diastolic window. This isolated segment was within the diastolic window i.e. right after the S2 sound (closure of aortic and pulmonic valves).
Blood flow during this period is maximal and therefore the turbulent acoustic signature is likely to be loud.

Various spectral based studies were used and they studied both normal and diseased patients. In all these studies it was shown that the diseased patients’ spectra had frequency peaks in the higher frequency ranges typically 300Hz and beyond. The different methods were employed to improve the signal to noise ratio. The sensitivity and specificity of these methods in predicting the existence of CAD varied, but some methods actually gave a specificity of 91% and sensitivity of 79% which suggested that acoustic methods were better than most other non-invasive methods to predict CAD.

![Spectral analysis of heart sounds: Normal (left) vs Diseased (right)](image)

Figure 1-7 shows a study where spectral analysis was applied to the heart sounds. The individual magnitude of the spectra was calculated using FFT and then averaged over 10 cardiac cycles. In the diseased patients more energy was observed in the 120-200Hz region [4]. Also this is consistent with earlier studies by [10] about the roll off at the break frequency.
1.4.1.1 Experimental Studies

Primitively speaking the detection of blockages can be seen as a study of a sudden constriction in a pipeline of uniform cross section. Such an occurrence is typically seen in arteries in the human body due to the buildup of cholesterol on the walls and hence a reduction in diameter. It is known that there is a drop in pressure across the blockage and this has serious ramifications on blood flow and the can cause heart attacks in humans.

1.4.1.2 Studies of pipelines and blockages

There have been studies of trying to detect blockages in circular pipes using vibration analysis [11]. The objective of that study was to correlate between the blockage levels and vibration signal. It is observed that when the fluid moves through the constriction the streamlines do tend to get closer causing an increase in fluid velocity and a decrease in pressure. As per Y. Kim et al [12] when fluid flows through an orifice, the high flow velocity and small constriction in the pipe will cause a drop in pressure. In this study an accelerometer was used at different locations to obtain the vibration response. The experimental set up is shown in the Figure 1-8. The vibration response was studied based on both the size of blockages as well as the locations of the accelerometers.
Mao Qing et al. [13] studied wall pressure fluctuations due to the presence of an orifice. The set up used to study the fluctuations has a centrifugal pump which drives the flow. The main test section is shown in Figure 1-9 with the orifice plate located at the center held together by a set of screws. The figure shows how this facility has the capability of having pressure transducers in both the longitudinal as well as in the circumferential directions. These are dynamic pressure transducers and are used to measure pressure upstream and downstream of the blockage albeit wall pressure fluctuations. The pressure transducers need to be mounted such that the flow field disturbances don’t influence the wall pressure fluctuations due to the presence of the orifice. The orifices used are machined in such a way that they are concentric with the pipeline and circular in shape as shown in Figure 1-9. The orifice plates are of varying sizes, fabricated as percentage of the unobstructed test section area.
Figure 1-9: Pipe design[13]

Figure 1-10: Orifice Plates[13]
1.5 Blockage Theory

Studies on human subjects have shown that when the arteries are almost completely blocked it is almost impossible to detect the blockage. The flow is so constricted that the flow is quiet and undetectable. Various methods have been used to try and detect orifice like blockages in pipelines. Adewumi et al [14] tried using the interaction between a pressure pulse in a pipe and the blockage within the pipe as a means for blockage detection as well as characterization. Kim et al [12] tried investigating the dynamic characteristics of orifices and blockages. In this experiment a single pipeline with orifices and blockages was investigated and the authors came up with frequency dependent and instantaneous inertia models to describe the dynamic flow behavior, specifically the kinetic pressure difference. Also from Adewumi’s study it is important to note that the method of pressure transients was used for early knowledge of both location as well as severity of blockage. To draw a parallel here, if an arterial blockage could be detected early it could make a significant difference.

![Figure 1-11: Expansion and contraction waves due to constriction][14]
Figure 1-11 from Adewumi’s study shows how the presence of a constriction or blockage causes pressure fluctuations without actually introducing forced excitation. It shows how the compression waves and expansions waves are formed at the inlet and exit of the blockage and how when the severity of the blockage increases more of the wave is reflected back which increases the pressure being recorded. The flow rate and kinetic energy are increased by the wave flowing through the blockage. At the outlet a similar sort of phenomenon occurs. The “pressure inlet response”[1] for different blockages is shown in the next figure and presents a strong case for having pressure transducers at multiple locations including the inlet in order to detect both the location as well as the extent of the blockage. Also there have been studies of how pulsating flow affects orifice flow meters [15].
1.6 Study of Pulsatile Flow

Stenosed flow is associated with disturbed pulsatile blood flow. The disturbances are found to be quite excessive during systole and more relaxed during diastole. Varghese, Frankel and Fischer [16] did a Direct Numerical simulation steady flow through stenosed tubes. They studied both axisymmetric and eccentric stenosis at inlet Reynolds numbers ranging from 500-1000. The DNS predicted a laminar post stenotic flow field immediately downstream of the stenosis. By introducing an eccentricity at the throat of the stenosis which is 5% of the pipe diameter the symmetry of the post stenotic flowfield is broken and at high enough Reynolds numbers such as 1000 jet breakdown occurs and about 5 tube diameters downstream flow transitions to turbulence. Wall shear stresses (WSS) were observed to be highest at the throat due to the accelerating fluid through the stenosis whereas WSS was low in the flow separation regions. Due
to this turbulent breakdown there are large spatial variations in WSS at the walls downstream of the stenosis. In this experiment the walls of the vessel were assumed to be rigid and the no slip condition was applied. [17]

Pulsatile jet flow was studied by [9] and the turbulence intensity spectra was studied. Orifice sizes were 50, 70, 80, 89% restriction of artery area wise. The power spectral density was computed both for the systolic as well as diastolic phases. It was computed by a program which incorporated a Fast Fourier Transform. The two major groups the experiment was divided into were the (i) stenosis size and variations in flow rate and (ii) pressure pulse shape variation. The frequency or pulse rate was varied from 50-110 bpm and systolic interval varied from 200-500ms. It was found that distal to the 89% orifice, flow was neither laminar nor turbulent but rather a disturbed flowfield, with turbulent jet features during systole and the dissipation of these disturbances during diastole.

1.7 The Pressure Pulse [18]
The pulse shape in the various arteries depends on the physiological and pathophysiological “conditions of the organism”[18]. There are different methods for pulse wave measurements depending on the type of measured pulse wave [18]. During the systole and diastole a certain amount of blood is ejected and each of the arteries and veins are affected by a pulse wave. Here 3 phenomena can be seen: increase in the blood pressure which is the pressure pulse, the blood flow which is the flow and a volume pulse [18]. The overall Pulse wave (PW) as mentioned earlier depends on both physiological condition which includes conditions such as the heart rate and the pathological conditions which include diseases such as atherosclerosis or diabetes as these affect the elasticity of the wall.
Blood circulation in the arteries can be seen as a mechanical system [18] composed of a piston and elastic hose. The movement of the piston causes a wave to propagate which causes three effects: a flow wave, a pressure wave, and a volume wave [18]. The pressure is a result of standing waves which occur due to the forward wave being reflected back from the periphery to the central vascular system [18]. When the artery becomes stiffer, the velocity of the pulse wave increases and the speed of the reflected wave correspondingly increases. This might cause the reflected wave to reach the aortic valve before it actually closes and hence there is an increase in blood pressure. In the ascending aorta, the pressure falls after reaching a maximum pressure with an incision at the end [18] of the diastole. This sort of shape is typical for the pressure pulse for arteries close to the heart. The pressure curves in arteries farther away are more rounded due to attenuation of the higher frequencies. Also due to vascular diseases, there is damping and a plasticity loss due to wall rigidity.

1.7.1 Pulsatile flow generation

Two of the mechanisms that could be used to simulate the pumping action of the heart are the cam follower and the slider crank. The cam follower is a mechanism in which the motor isn’t the only one controlling the pumping action but also the profile of the cam itself. Using a software such as Dynacam [18] the shape can be determined taking into consideration the required force on the piston and the displacement with respect to time needed. The shape required for a pumping action close to the heart is an oval shape with bends and concavities. The concave part allows for a pause between pumps as is the case physiologically. The frequency can be controlled by the motor and hence the desired type of flow wave can be produced. The slider crank mechanism on the other hand is very much like the one used in car engines. The piston is directly connected to the flywheel via a connecting rod, hence the cam has a circular shape. The advantage of this system is that friction and wearing aren’t a problem as in the cam follower
where wear can cause an alteration in the path. Also most importantly control is fully controlled by the motor and here kinematics is very important to get the right length angles and hence the right kind of wave. A slider crank mechanism in constant motion would only produce a sine wave which is not what the normal human body produces.

Additionally a motion controlling device needs to be included in order to produce the sort of wave required.

1.8 Stenotic Hemodynamics
A stenosis is characterized by a narrowing of the arterial wall and a protrusion which basically causes a converging diverging morphology. For the converging region the Bernoulli equation is applicable but for the stenosed region the Poiseulle’s law describes the flow better. However the post stenotic region is characterized by flow separation and the flow is turbulent and there are viscous losses due to flow separation in that region. Also due to the protrusions the increase in resistance causes a pressure drop along the flow [19]. The pressure loss can be related to the flow Q with semi empirical relations.

Figure 1-13: Flow through a stenotic region[19]

\[
\Delta P = a_1 Q + a_2 Q^2
\]
It is seen here that the pressure drop shown in the figure is related to flow rate $Q$ as a quadratic function. The first constant $a_1$ is dependent on the viscosity of the fluid, the length of the stenosed area and the area of the stenosis. This first term represents the viscous losses. Meanwhile $a_2$ is dependent on the area of the stenosis, the area of the unobstructed vessel, the density and a constant which is dependent on the geometry of the stenosis.

Blustein studied the effect of varying degrees of stenosis on the flow characteristics of turbulent pulsatile flow through heart valves[20]. This study was used as a guideline for valve design. Previous studies have established well the fact that turbulence levels increase with increasing severity of stenosis. But there is also the occurrence of intermittent turbulence and at a certain time in the cardiac cycle that needs addressing. In vivo experiments are difficult to conduct to understand the fluid mechanics of cardiovascular flow situations, but in vitro setups have been developed. Here it must be mentioned that a comparative study of both steady and pulsatile flow was done and the rms levels in the steady state were upper bounds for that measured in pulsatile flow[21] [22]. These studies show that major portions of the pulsatile flow cycle cannot be inferred from steady flow experiments. The steady flow assumes highest rms levels at peak flow whereas in the case of pulsatile flow the magnitude of the fluctuating velocities depends on the systolic phase.

1.8.1 Properties of Blood

Before looking into the behavior and how the pressure time traces look it is paramount to understand the circulatory system. The heart acts like a pump and creates pressure gradients. It is this gradient that transports blood through the various vessels. The percentage of blood flowing through the various vessels is unevenly distributed with a majority flowing through the veins. About 13% flows thorough the major arteries and the rest through arterioles and capillaries.
The Aorta is the largest vessel with about a cross sectional area of 2.5 sq cm. The average velocity of the blood flowing through it is about 33 cm/s. At normal heart functioning the arterial pressure typically fluctuates between systolic pressure of 120 mmHg and 80 mmHg diastolic pressure. It must also be remembered that the blood vessels are compliant. During the systole phase when there is ejection of blood from the heart into the aorta a pressure pulse is transmitted through the system of arteries. The traveling velocity of the pulse in the aorta is about 3 to 5 m/s. The following figure shows variation in pressure and velocity of the pulse in the various vessels through which the blood flows. Figure 1-14 shows the pressure waveforms in arteries.

![Figure 1-14: Pressure waveforms in various arteries][23]
1.9 Fluid Mechanics
Studies on the fluid mechanics of the aortic stenosis by [24] [25] tried showing the relationship between the flow and the pressure drop across a stenosis, both in the case of steady as well as unsteady flow. The mechanical set up of having a sudden constriction in a pipeline is comparable to an aortic stenosis. Due to the sudden contraction and then expansion energy is dissipated due to the turbulent mixing and loss of mechanical energy as heat. In a real aortic stenosis more energy could be lost due to post stenotic dilation. In case the obstruction is smooth and gradual the pressure recovers better and the losses are lower. This study tries to find a relation between the flow and the pressure recovery downstream of the stenosis. Both steady and unsteady experiments were conducted. The tube was assumed to be rigid.

![Flow Pattern in Aortic Stenosis](image)

Figure 1-15: Flow Pattern in Aortic Stenosis[24]

The flow pattern in an aortic stenosis can be seen in the figure above. Once the fluid exits, the expansion the region near the wall develops a recirculation region which in the figure is denoted
by ‘R’ The core of the fluid moving centrally remains laminar for some distance. Then at point A the flow reattaches to the wall and downstream of this the flow is typically turbulent. It must be noted that the transition to turbulence after a sudden expansion happens at relatively low Reynolds numbers.

More important to the current experiment though is the study which deals with unsteady flow that is pulsating flow in this case. The objective is to see whether under the pulsating conditions the pressure recovers after the stenosis especially because if it does rise appreciably along the aorta, then the positioning of the pressure transducer to pick up measurements becomes paramount. Previous studies even gave a formula to calculate valve area but downstream pressure recovery wasn’t taken into account then. Modeling parameters used during the steady tests were the area ratios and Reynolds number. Additionally in the pulsating case, using dimensional analysis, two more parameters were introduced, ratio of flow rates $Q_p/Q$ that is ratio of peak flow rate and mean flow rate and the Strouhal number $St=fd/U$. The Strouhal number basically is a ratio of the local to convective acceleration.

<table>
<thead>
<tr>
<th>Table 1. Parameter values for man</th>
</tr>
</thead>
<tbody>
<tr>
<td>$A_2/A_3$</td>
</tr>
<tr>
<td>------------</td>
</tr>
<tr>
<td>1. Normal</td>
</tr>
<tr>
<td>2. Mild Stenosis</td>
</tr>
<tr>
<td>3. Severe Stenosis</td>
</tr>
</tbody>
</table>

Where: $f$ = heart rate and $t = duration$ of ejection (s). Note: $Q = mean$ flow based on ejection time.

Table 1-2: Parameter values[25]
2 Facility Design

Figure 2-1: Facility Schematic

2.1 Objective

Figure 2-1 shows the schematic of the facility. The designed facility has the capability to simulate both steady as well as pulsatile flow that is observed in the human aorta. In the current study, we are interested in studying the pulsatile flow through orifice like
blockages which basically simulates a stenosed artery with the heart beating at the upstream end. The blockage is an orifice plate and is axisymmetric [1]. Though stenoses in the arteries do not form in this symmetric fashion this blockage serves as a means to study a quantifiable blockage. Also since the flow in the artery is not steady, this facility has the capability to study pulsating flow as well and the results obtained can be compared to the steady results [1]. Besides the varying size of blockage, flow at multiple velocities can be studied via changing pressure head that drives the flow. Depending on the state of the heart, the rate or frequency of the pulsating flow changes and this facility has the ability to change the frequency of pulsation. In summation the effect of three important parameters is studied, varying pressure/velocity the degree of constriction, and the frequency of pulsation. In order to quantify the impact of the blockage, surface pressure measurements and flow visualization is done. In addition to these capabilities it is interesting to note that with this particular experiment the duty cycle of pulsation can be varied and the location of pulsation can be varied to either upstream or downstream of the blockage. This serves for an interesting set of comparisons while studying the spectra and pressure characteristics.

2.2 Test Section:

![Image](image.png)  
**Figure 2-2: Test Section**
Before delving into the facility set up, the test section needs to be understood, as this is where the pressure measurements are made. The test section is a transparent tube made of Plexiglas and consists of two identical halves. One end of each section has a flange and the blockage is inserted between the two flanges. As can be seen in Figure 2-2 the two halves are attached in the center at flange end. The brass fittings are used to mount the pressure transducers. To measure the pressure at any of the six locations, a pressure transducer is mounted in one port and the rest of the ports are closed with these brass fittings so that the water does not leak while the experiment is being conducted. Figure 2-3 is a schematic of the test section and blockage and shows how ΔP is calculated. The test section is long enough for turbulent flow to fully develop but not laminar flow. According to computational work done previously, the minimum entrance length for fully developed turbulent flow is given by Equation 2.1. [1]

\[ L = 0.693R e^{\frac{1}{2}}D \]  

(2.1)
Since the maximum pressure head that will be seen in the BT is four inches, the Reynolds number can be calculated from Equation 2.2 shown below

\[ Re = \frac{\rho UD}{\mu} \]  \hspace{1cm} (2.2)

to be equal to approximately 35,790. This means that according to Equation 2.1, the minimum entry length for fully developed flow is about 8 ¾ in, which is more than satisfied by the test section [1]. The inner diameter of the test section is 1” and compares to the diameter of the aorta.

Figure 2-4 below shows the flange end of half the test section. The orifice plate or blockage is fitted between the two flanges with the help of a seal which makes the fit water tight and prevents any leakage during the experiment. In the current setup the O-ring has been done away with and instead the entire flange cross-section has a water tight seal.

\[ \text{Figure 2-4: Flange End, Test Section Half [1]} \]
2.3 Blockage

![Blockage schematic](image)

Figure 2-5: Blockage schematic

Figure 2-5 shows the schematic of the blockage inside the test section as well as the various blockages that were fabricated. This blockage is similar to the one fabricated and studied in Qing M et al [13]. Figure 1-10 shows the fabricated orifice plates and are similar to the blockages fabricated for the current setup. The blockage is at the center and is fitted at the flange end between the two test section halves [1]. This is similar to how the set-up is in [13] as shown in Figure 1-9.

2.4 Facility set up
Figure 2-6 shows the schematic of both the previous facility and the current facility. The current facility consists of two tanks, upstream and downstream tanks. As the name suggests the
upstream tank is where the flow begins. The two tanks are connected by the transparent acrylic test section mentioned in Section 1.2. From the Upstream tank the fluid flows through the test section in which the stenosis is located right at the center. Flow through this stenosis is studied under various flow conditions. The flow from the test section empties into the downstream tank. The Upstream tank has an inlet valve which controls the inlet flow rate. A hose from a faucet is directly connected to the inlet valve of the upstream tank and supplies a continuous inflow of water. It must be mentioned that a constant head is maintained in this upstream tank and it is this head that drives the flow into the test section. Before the experiment is started the test section must be full in addition to both the upstream and downstream tanks being held at constant heads. It is the height difference between the two tanks that drives a continuous flow. Water from both tanks drain into a third tank, the drain tank. The flow into the test section from the upstream tank and the flow to the downstream tank from the test section are controlled by two identical 1” ball valves.
2.5 Tank Design

In the present set up the upstream tank is a 23 gallon 18”x18”x18”, 5/8” thick polyethylene tank. It must be noted in the previous setup, on the upstream side of the blockage there were two tanks namely the baffle tank and the head tank. It was the baffle tank that was connected to the test section and on top of the baffle tank lay the head tank which was a sort of reservoir and was connected to the baffle tank via a ball valve. Controlled flow from the head tank into the baffle tank could be achieved this way and a constant head was maintained in the baffle tank while conducting the experiment. In the current setup the head tank has been removed and instead of...
two there is just one tank viz the Upstream tank. It is the constant head in this tank that drives the flow through the test section and into the downstream tank. This tank now does not just play the role that the baffle tank played previously but in addition it has the whole pulsating set up in it.

The Upstream tank has 3 holes drilled into it where bulk heads are fitted. The inlet and the outlet to the test section are fitted with 1” bulk head fittings. In addition a third 1/2” hole with a bulk head is fitted at the bottom of the tank. This outlet is to drain excess water coming into the tank to maintain the head at a constant level. The drain pipe from the head tank has a valve to control the rate of outflow. Inserted into this 1/2” bulk head in the interior of the tank are stand tubes of various heights to maintain the head at the desired level. As mentioned earlier this tank has the complete set up to pulse the flow and that pulsating capability is explained in greater detail later.

On the downstream side the test section empties into a similar polyethylene tank 18”x24”x12”. This was the only pulsating location in the previous setup. It currently still has the ability to pulse the flow hence the new facility has the capability to pulsate the flow both upstream and downstream. This tank is shorter in height because the head needs to be maintained at a much lower height. This tank too has 3 holes drilled into it with bulk head fittings. The 1” inlet has the test section entering into it. Two 1/2 inch holes are made in order to drain the tank and keep the head constant. The hole at the bottom has a pipe with a valve attached so that the draining flow rate can be regulated as desired. Also on the opposite wall to the inlet is another 1/2” hole into which is fitted a stand tube that drains out the water and maintains the incoming flow rate equal to the outgoing flow rate to keep the water height constant in the downstream tank. All the water that drains out of both the upstream and downstream tank goes into a third drain tank from which the water goes out into a drain. At the moment the water is not being reused, but an option in the
future could be setting up a pump to recirculate the water from the downstream tank back to the upstream tank.

2.6 Flow Pulsing
The pulsations in the flow are introduced via a stepper motor rotating a wheel. The stepper motor is mounted on a platform fixed across the tank 14” above the base. This is one of the primary reasons to have used an upstream tank of height 18” as the motor cannot be allowed to come in contact with the water. Connected to the motor is an aluminum shaft and to the other end of the shaft is connected the wheel. The wheel has equally spaced holes and when the motor is powered up the wheel spins with the help of the shaft. The holes periodically go over the inlet causing an opening and closing which creates the pulsating effect. The amount of time the inlet needs to be open and the speed at which the wheel needs to rotate is determined by the actual cardiac cycle of the human body. Below are two figures that show the motor-shaft-wheel set up inside the upstream tank. Figure 2-7 shows a top view of the upstream tank where the motor and wheel can be seen. Figure 2-8 is a close up of the rotating wheel set up. The acrylic housing is designed such that it holds the wheel shaft assembly in place. Also the housing has a cavity or hole in which the inlet pipe sits. As can be seen in the figure the wheel on rotation passes over this inlet.
2.6.1 Rotating wheel design

The wheel design i.e. the number of holes its sizing and the space between them is decided by the durations of systole and diastole. In the current study being conducted the frequencies of the pulse being studied are 1Hz and 2Hz corresponding to the heart beat of humans at rest and humans doing vigorous exercise respectively. Also the systole diastole ratios differ based on the severity of stenosis i.e. whether it is a diseased artery or normal. V Gemignani et al [6] conducted studies using non-invasive methods and with the help of the ECG signal and acceleration signal that was obtained from the positioning of an accelerometer sensor, plotted systole/diastole ratios as a function of heart rate. Figure 2-10 and Figure 2-11 show the plots. It is observed from the plot, taking into the consideration the error bars, that a S/D=0.52 at 60 bpm and S/D=0.85 at 120 bpm are reasonable assumptions to design the wheel. For the 1Hz or 60bpm, case it was decided that two diametrically opposite circular holes of diameter 1.375” be machined to match S/D=0.52. In the 2 Hz or 120bpm case 3 equally spaced holes of diameter 1.125” are machined to create an S/D ratio of 0.85. In the already existing wheel design, 4 holes
were machined 90 degrees apart and the S/D ratio is slightly greater than the desired S/D ratio, hence a new design was generated and two new wheels machined. Figure 2-9 shows the same.

Figure 2-9: Wheel designs based on duty cycle

Figure 2-10: Systole and Diastole timings[6]  Figure 2-11: Systole/Diastole or S/D ratio vs rate[6]

Corresponding to the systole period is when the hole in the wheel slowly starts lining up over the inlet and allows the water from the upstream tank to flow into the test section. This periodic
opening and closing of the inlet (systole and diastole) results in the pulsating flow in the facility. One of the objectives of this experiment is to be able to match the pressure profiles obtained from the pressure measurements made upstream of the blockage to the actual human aortic pressure profile. This is the reason why the S/D ratio and the wheel design is of great importance while trying to predict the extent of the blockage by looking at the FFT spectra and the pressure time traces.

2.6.2 Geometry and Design of wheel
The previous facility [1] had a wheel that had 4 symmetric holes 90 deg apart. This facilitated the flow, no flow design and caused the pulsation in the flow. Every time the hole passes over the inlet there is flow and once the solid part of the wheel moves over the inlet the flow is blocked. In terms of the heart pumping blood through the arteries this sort of flow corresponds to the systole and diastole phases of the duty cycle. Depending upon the heart rate and the state of the artery, the time of the systole and diastole or more importantly the S/D ratio is affected. Here we look at how the design was conceived to match the timings and S/D ratios to suit both the heart rate and the state of the artery (healthy vs diseased). The theory behind the design is the simple concept of intersecting circles.
The two pictures in Figure 2-12 show the wheel individually and when completely set up i.e. connected to the stepper motor via the aluminum shaft. ‘R’ here is the distance between the center of the wheel and the center of any of the holes on the wheel. In other words the locus of the center of the hole is a circle of radius ‘R’. Let the radii of the hole and the inlet (pipe) over which the hole on the wheel traverses be R1 and R2, respectively. Since the wheel is spinning at a constant rate of either 1Hz or 2Hz, the hole keeps travelling over the hole at a constant rate. In this particular design problem, the radius of the hole, the number of holes and hence the spacing of the holes needs to be calculated such that the S/D ratios and systole diastole times mentioned in the literature can be achieved. Hence, at the basic level, this problem is broken down to the intersection of two circles viz the hole and the inlet and the time it takes for the area to open and close. The time the inlet is open is the systole time and the time it is closed is the diastole time.
Figure 2-13 shows the basic schematic of the hole and the inlet at a certain instant in time. The overlapping portion is the area that is open for the flow to take place at any point of time. Using basic geometric relations the area of the sector can be calculated using areas of triangles.

\[ A_{Overlap} = (A_{SecABC} - A_{TrABC}) - (A_{SecEBC} - A_{TrEBC}) \]  \hspace{1cm} (2.3)

where \( A_{Overlap} \) is the area of the overlapping portion, \( A_{Sec} \) is area of the sector and \( A_{Tr} \) is area of the triangle.

\[ A_{SecABC} = 0.5 \times R_1^2 \times (2 \times \theta_1) \]  \hspace{1cm} (2.4)

\[ A_{TrABC} = 0.5 \times R_1^2 \times \sin(2 \times \theta_1) \]  \hspace{1cm} (2.5)

Similarly equations can be written for sector and triangle EBC. Based on the angular velocity of the wheel, the spacing and size of the holes is decided. It must be remembered however that the
radius of the locus of the center of the hole and the distance between the center of the wheel and the inlet is equal, hence when the hole and the inlet are concentric maximum discharge occurs. To draw a parallel, this is equivalent to the aortic valve being completely open during the systole phase. The diameter of the hole and the number of holes is decided upon by using the S/D ratios from literature.

3 Experimental Setup [1]

The objective with the current facility is to simulate the heart beat and the flow of blood in the human body as closely as possible. To understand parametric studies are done. The effect of Head, Blockage, Frequency and duty cycle is studied. Pressure measurements are taken and pressure drops measured across the blockage.

3.1 Equipment Required
As mentioned earlier in section 2.1 the objective of this experiment is to conduct surface pressure measurements and flow visualization to study the impact of pulsating flow on blockages. Surface pressure measurements are conducted using a pressure transducer by picking up the pressure at the extreme upstream and extreme downstream port.

Firstly, the flow needs to be pulsatile. In order to achieve this the internal motor-wheel set up has been fabricated and discussed in Section 2.4. A Vexta PK266-02A stepper motor is used to rotate the wheel via an aluminum shaft. This is identical to the motor that had already been installed in the downstream tank previously. The stepper motor is connected to a stepper motor controller which in turn is connected to the communicator. This allows the motor to communicate with the computer via the controlling software COSMOS. The motor needs to be setup appropriately to
move the wheel at the right speed which then translates to the right frequency. The software has two options: Absolute Move and Relative Move. Select Relative Move and to achieve the right pulsation frequency since it is a stepper motor the number of the steps and the steps per second need to be set to the right value. [1]

The next piece of equipment is the pressure transducer. A Validyne DP-15 transducer is used to measure pressure upstream and/or downstream of the flow blockage. The DP -15 transducer is screwed into the pressure tap/port on the test section. The transducer is connected to a Validyne CD-23 display which displays readings in terms of voltage. The accuracy of the display can be set to one hundredth of a volt, its range being -5 to 5 V. The National Instruments DAQ system NI BNC-2110 interfaces the pressure transducer to the computer. Once the CD-23 is connected to the DAQ and the DAQ to the computer, the computer displays the voltage values of the pressure picked up by the Validyne DP-15 pressure transducer. LABVIEW talks to the DAQ via the Labview VIs.

3.2 Pressure Measurements
Single port pressure measurements are taken at Port A (far Upstream port) and Port F( far downstream port). As can be seen in Figure 2.5.1 there are 6 brass fittings that screw into the test section. When taking a pressure measurement the Validyne DP-15 is screwed into either of Ports A or F with the help of the new brass fitting. The old and the new fittings are shown in Figure 3-1.
The longer, old one was used previously but to record the pressure more accurately the smaller, new fitting is used to bring the pressure transducer closer to the flow. Since this is a wet-wet differential pressure transducer there is a positive and negative side on either side of the diaphragm. The positive end via the brass fitting is screwed into the port at which the pressure needs to be recorded. The negative end has another brass fitting to which a 1/8” flexible tubing is connected. This tube is submerged in a bottle of water to keep the signal near zero. Once the pressure transducer is set up, the CD-23 display, the labview VI acquirepressure.vi and the stepper motor controller must all be connected and ready to run.
3.3 Conducting Pressure Measurements
Before running an experiment, parameters that will be studied need to be set. The parameters that can be controlled are Head, Frequency and extent of blockage. The 4 heads that can be achieved in this experiment are 1in, 2in, 3in and 4in of H2O. For each of the different heads there are different stand tubes. Once a certain head and a specific blockage is chosen, the test section is allowed to fill up, until the head in both the upstream and downstream tanks are brought to a constant level. Once this is done, both the valves upstream and downstream of the blockage must be kept open to allow the flow through, and the flow rate adjusted to such a level that the head remains constant. Once the facility is allowed to reach this steady state, the pressure transducer, with the (+) end screwed into either of Port A or Port F and the (-) end saturated in the water bottle, is zeroed. The height at which the water bottle is kept adjusts the zeroing on the transducer and when it is close enough to zero the CD display box can be manually zeroed. It is important to ensure that the holes on the wheel line up right over the inlet for the above to be achieved. Once the facility is set up in this state it is time to use the stepper motor controller and run the stepper motor. The Velmex controller software is used and the ‘relative motion’ button needs to be clicked. Assuming flow is pulsing at 1Hz, the motor is set to a value of 100 steps per second. This ensures that at every one second interval a hole on the wheel traverses over the inlet. The number of steps is set to 10500 so that 100 seconds of data can be recorded. Once the motor is set in motion, the wheel starts spinning and the flow is periodically blocked and let through. The pressure transducer starts measuring fluctuating pressure due to the stepper motor rotation. Simultaneously, when the stepper motor is set in action, the Labview program is initiated. This facilitates the data acquisition and the voltage data output is recorded in the form of a text file. When running the Labview program the parameters need to be set exactly to those that are being run. For example if 4in Head, 65% blockage and 1Hz frequency case is being run
then these exact parameters need to be selected in the program. Also the location of the transducer needs to be mentioned, whether it is Port A or Port F i.e. whether the pressure is being recorded upstream of the stenosis or downstream of the stenosis. In addition since this facility has pulsing capabilities both upstream and downstream the location of pulsing needs to be selected too.

3.3.1 Problems encountered

When the experiment is conducted there are a few problems that the experimentalist faces. Some of them are listed below.

1. **Cavitation**: Due to certain flaws in the design of the test section and connecting pipes bubbles are sometimes formed, especially in the case of the 85% and 96% blockages. In order to minimize this, the valves need to be opened and closed in quick succession while one of the pressure taps is unscrewed so that the air bubble escapes out of the pressure port. Priming action is performed by trying to suck water out of the test section through the port while there is flow. This way the bubbles can be minimized and hence any errors in pressure readings can be reduced.

2. **Rotation of Wheel**: Another common problem is the distance travelled by the wheel. If the bolt holding the wheel and the shaft is slightly loose, a grating noise can be heard and more significantly the time period of rotation and hence the frequency is slightly off from the desired value. To avoid this the screw needs to be tightened properly.

3. **Leakage**: While measuring pressure make sure that there is no water leakage through the pressure port from below the transducer. Water can also leak if the blockage isn’t fastened and screwed tightly enough between the flanges with the rubber water tight seals. It may inadvertently cause air bubbles in the set up.
4. **Head Control**: Controlling head is another challenge especially with the lower blockages as flow rate is quite high. The inlet valve and the faucet need to be open just the right amount to maintain the flow rate accurate and constant.

5. **Equipment safety**: Most importantly never let the motor or the wiring, come in contact with the water as this may permanently damage the stepper motor.

3.4 **Changes to set up and changing Parameters**

Before going through the major changes that were affected in the setup it is important to first understand the changing of parameters when the experiment is being run. The 3 major parameters that are studied are:

1. **Head**: In this set up the head can be set to either of 4 different values: 1”, 2”, 3”, 4”. As the head changes the flow rate and the velocity changes. In comparison to the previous facility the only change, is the way in which head is controlled. In the current set up stand tubes are used instead of a valve.

2. **Blockage size**: The different blockage sizes are:

<table>
<thead>
<tr>
<th>Blockage</th>
<th>0%</th>
<th>30%</th>
<th>65%</th>
<th>85%</th>
<th>96%</th>
</tr>
</thead>
<tbody>
<tr>
<td>Diameter(in)</td>
<td>1</td>
<td>0.836</td>
<td>0.592</td>
<td>0.388</td>
<td>0.20</td>
</tr>
<tr>
<td>Area(in^2)</td>
<td>0.785</td>
<td>0.549</td>
<td>0.275</td>
<td>0.118</td>
<td>0.031</td>
</tr>
</tbody>
</table>

Compared to the previous facility the 85% has been additionally fabricated

3. **Frequency**: The frequencies studied in this particular experiment are 1Hz and 2Hz.

In the current facility the experimentalist can change the duty cycle or the Systole/Diastole(S/D) ratio by changing the wheel. Also as has been mentioned previously the fitting on the transducer
has been changed to mount the transducer closer to the flow for better pick up of the pressure signal. With the same above objective in mind the size of the port into which the pressure transducer is screwed in via the mount was widened progressively as can be seen in the figure until it was decided that the effect of widening the hole doesn’t further enhance or aid the ability of the transducer to pick up the signal. Other than these changes the one visible change is the removal of the former head tank which was already mentioned in Section 2.3. Figure 3-2 shows some of the changes mentioned.

![Figure 3-2: Changes made to the Experiment](image)

### 3.5 Flow Visualization

Pressure measurements and plots of pressure drops can give only a certain amount of information. In order to understand the flow better and to co-relate the pressure drops to both the pulsation frequency as well as the extent of the blockage, visualization of flow is a very helpful and important tool. There are various methods typically used in previous studies. PIV and LDV are common and effective methods to analyze the flow quantitatively. In this particular
experiment, dye injection is used and this is more a qualitative method. The objective of this is to see the flow pulsing and to observe the fluid flowing through the blockage with the help of the dye acting as tracer. Depending on the size of the blockage and frequency of pulsation, the flow behaves differently and the presence of vortices and turbulence can hopefully be visualized.

The dye used was red food coloring. It is injected into the tube via a 1/8in stainless tube through the same port that was used to mount the pressure transducer. Once the experiment is running, with the wheel spinning, the flow visualization experiment is started. The flow which has been injected with dye is visualized using a video camera. The movies are then imported into MATLAB and the appropriate frames are extracted using the getframe command. These still frames show the features of the flow that can be then analyzed. Figure 3-3 shows the initial field of view for the flow visualization.

![Figure 3-3: Flow Visualization Arrangement](image)
3.6 Conducting Flow Visualization
In 3.5 the method of flow visualization has been mentioned. Here the method of conducting flow visualization is explained. As explained in Section 3.5 a bottle of red food coloring is needed to be used as a dye. The other equipment used is a 1/8” flexible transparent tubing a ½” brass or stainless steel reducer nipple to screw into the far fore pressure port or Port A, a stainless steel 1/8” tube, a tripod stand and a video camera. Firstly the nipple needs to be fitted with 1/8” stainless steel tube. This tube is bent at right angles to the vertical such that it enters the test section in the direction of flow at the centerline of the test section. Figure 3-3 shows the field of view along with the labelled parts. The test section is already full and the water level in the tank is at the required head. The tube is fitted tightly and inserted with the bent portion entering the test section. The other end of the stainless tube protrudes vertically through the nipple to the outside and is locked using a fitting. This end is fitted with the flexible tubing and the other end of the flexible tubing goes into the bottle of dye. The bottle of dye must be placed carefully so as to make sure that it does not topple over. Before these connections are made though it must be remembered that the dye has to be first bled into the tube and then connected to the stainless steel tube. This way the air is driven out of the tube. In order to check whether the dye is flowing hold the bottle up at a higher head than the water and let the dye bleed into the test section. Once this happens the experimentalist can be sure that air has been driven out and that the injection can take place. Make sure that the dye is placed at a level where the pressure head is equal otherwise the water from the test section will run back into the bottle. Alternatively the flexible tubing can be bent and kept pinched by tying it up with a small tie or rubber band so that there is no fluid movement either way.
As close to the test section as possible the video camera is mounted on a tripod stand. The cart on which the tanks sit has an open portion in the center and through this portion the tripod can be adjusted. The camera is placed such that just the 6 ports are visible. It is necessary for the camera to be pointing downwards at about a 45 degree angle to be able to capture the flow well. To be able to visualize the flow as clearly as possible the immediate background behind the test section is made white by placing an A4 sided sheet or any other white background. The surrounding region farther away from the test section is covered with black paper. This is done so that there isn’t too much light which can cause reflections off the acrylic test section and hinder the clarity of the image. There must be enough light though for the video camera to be able to capture the image clearly.

Figure 3-4: Flow Visualization in Progress

Once the setup is ready the faucet can be opened and the two valves at either end of the test section should be opened to let the flow through. The camera is then switched on and the recording begins. As soon as the recording begins the connected stepper motor should be
powered up and once the wheel starts spinning the facility is ready for dye injection. The tie or the rubber band can be removed and slowly the bottle should be raised to a higher level. The dye will slowly bleed and enter the test section. After about 10 to 15 seconds when the dye mixes almost completely with the flowing water the experiment can be stopped. Thus the method of experimentation is explained. The next two sections will discuss the results obtained by such experiments and the results will be discussed in light of some of the findings of experimental analyses. Figure 3-4 shows the flow visualization process in progress.

### 3.7 Test Matrix
Table 3-1 shows a test matrix for all the cases that were run. Table 3-2 shows those cases for which flow visualization was done. The two tables give an entire picture of the types of experiments and number of cases run.

#### 3.7.1 Upstream Pulsing Pressure Matrix

<table>
<thead>
<tr>
<th>Blockage</th>
<th>0%</th>
<th>30%</th>
<th>65%</th>
<th>85%</th>
<th>96%</th>
</tr>
</thead>
<tbody>
<tr>
<td>Head(in)</td>
<td>1%</td>
<td>30%</td>
<td>65%</td>
<td>85%</td>
<td>96%</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>Old Wheel</th>
<th>1 Hz</th>
<th>2 Hz</th>
<th>1 Hz</th>
<th>2 Hz</th>
<th>1 Hz</th>
<th>2 Hz</th>
<th>1 Hz</th>
<th>2 Hz</th>
<th>1 Hz</th>
<th>2 Hz</th>
</tr>
</thead>
<tbody>
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<td>No</td>
<td>Yes</td>
<td>No</td>
<td>Yes</td>
<td>No</td>
<td>Yes</td>
<td>No</td>
<td>Yes</td>
<td>No</td>
</tr>
<tr>
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<td>Yes</td>
<td>Yes</td>
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</tr>
<tr>
<td>3</td>
<td>Yes</td>
<td>Yes</td>
<td>Yes</td>
<td>Yes</td>
<td>Yes</td>
<td>Yes</td>
<td>Yes</td>
<td>Yes</td>
<td>Yes</td>
<td>Yes</td>
</tr>
</tbody>
</table>

49
<table>
<thead>
<tr>
<th>New Wheel</th>
<th>4</th>
<th>1Hz</th>
<th>Yes</th>
<th>Yes</th>
<th>Yes</th>
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</tr>
</thead>
<tbody>
<tr>
<td></td>
<td></td>
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<td>Yes</td>
</tr>
<tr>
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<td>1Hz</td>
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</tr>
<tr>
<td></td>
<td></td>
<td>2Hz</td>
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<td>Yes</td>
<td>Yes</td>
</tr>
<tr>
<td>2</td>
<td>1Hz</td>
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</tr>
<tr>
<td></td>
<td></td>
<td>2Hz</td>
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<td>Yes</td>
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<td>Yes</td>
</tr>
<tr>
<td>3</td>
<td>1Hz</td>
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<td>Yes</td>
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<td>Yes</td>
</tr>
<tr>
<td></td>
<td></td>
<td>2Hz</td>
<td>No</td>
<td>No</td>
<td>Yes</td>
<td>Yes</td>
<td>Yes</td>
</tr>
<tr>
<td>4</td>
<td>1Hz</td>
<td>No</td>
<td>Yes</td>
<td>Yes</td>
<td>Yes</td>
<td>Yes</td>
<td>Yes</td>
</tr>
<tr>
<td></td>
<td></td>
<td>2Hz</td>
<td>No</td>
<td>No</td>
<td>Yes</td>
<td>Yes</td>
<td>Yes</td>
</tr>
</tbody>
</table>

Table 3-1: Pressure Measurement Test Matrix
### 3.7.2 Flow Visualization Matrix

<table>
<thead>
<tr>
<th>Percentage (%)</th>
<th>Flow Configuration</th>
</tr>
</thead>
<tbody>
<tr>
<td>0%</td>
<td>3in, 1Hz, New Wheel</td>
</tr>
<tr>
<td>30%</td>
<td>3in, 1Hz, New Wheel, Upstream Pulsing</td>
</tr>
<tr>
<td>65%</td>
<td>3in, 1Hz, New Wheel, Upstream Pulsing</td>
</tr>
<tr>
<td>85%</td>
<td>3in, 1Hz, Old Wheel Pulsing</td>
</tr>
<tr>
<td></td>
<td>3in, 1Hz, Downstream Pulsing</td>
</tr>
<tr>
<td></td>
<td>3in, 1Hz, New Wheel, Upstream Pulsing</td>
</tr>
<tr>
<td>96%</td>
<td>3in, 1Hz, New Wheel, Upstream Pulsing</td>
</tr>
</tbody>
</table>

*Table 3-2: Flow Visualization Test matrix*

## 4 Results

### 4.1 Pulsating Set up

With the Vexta PK266-02A the motor can be commanded to move a specific number of steps per second for a prescribed amount of time. As described earlier the motor is connected to a shaft which is connected to a wheel at the other end. It is the movement of this wheel that causes flow being blocked and allowed through. This pulsation effect through the blockage is trying to
simulate the beating of the heart and with the FFT spectra and pressure time traces help quantify the physics of the flow through various blockages at different constant pressure heads. The pulsation frequencies studied are 1Hz and 2Hz. These typically correspond to the rate at which the heart beats when the body is at rest and during vigorous exercise conditions respectively.

4.2 Upstream Pulsing
The facility can be pulsed upstream as well as downstream of the blockage. First, the case of upstream pulsing is looked at in detail and quantified using the pressure time traces taken upstream and downstream of the blockage.

![Pressure Time Trace](image)

**Figure 4-1: Pressure Time Trace**

4.2.1 Pressure Time Traces
Figure 4-1 shows a sample case of the fluctuating pressure in the time domain. The red time traces is the unfiltered signal and the blue time trace is the filtered signal using a 10 Hz low pass filter. By removing the high frequency components in the signal, the low frequency components are more discernible. The time traces show a series of crests and trough with a periodicity of 1 sec which is consistent with the 1 Hz pulsing frequency. One important observation is that the pressure pulse is on the negative side of the axis. When the experiment is conducted, the pressure
transducer needs to be zeroed at a particular state of flow. In this experiment the transducer is zeroed at steady state that is when the valves are completely open and the head is constant. During upstream pulsing the wheel traverses over the inlet in the upstream head tank and this causes a periodic opening and closing of the inlet. The zero pressure corresponds to when the inlet is completely open and the pressure is negative because the inlet closes. When the inlet gets blocked the effective head on the upstream side ceases to exist as this head is not driving the flow anymore. Additionally due to the way the set-up is fabricated there is a small head at the exit which can cause a backflow. These two factors explain the negative pressures observed when the facility is pulsed upstream. A typical pressure time trace looks like the one seen in Figure 4-2. It shows how the pressure pulse lines up when measured at both ports A and F. The pressures recorded at A are higher than that recorded at port F. This suggests that there is a pressure drop across the blockage.
This time trace can be compared to the aortic pressure time trace shown in Figure 4-3. One of the main objectives in this thesis is to match the experiments to the aortic pressure curve. For that purpose, the duty cycle and systole diastole ratios are controlled with different wheel designs.
4.2.2 FFT

Figure 4-4 shows the FFT as well as the corresponding pressure time trace. The particular case being considered here is: 3in head, 85% blockage and 1Hz frequency. Figure 4-4 shows the FFT spectra obtained from measuring pressures at two different locations. The higher amplitude spectrum is for the pressure measured at the port farthest upstream of the blockage (Port A).

The lower amplitude spectrum corresponds to the measurement made at the far downstream location (Port F). The amplitude on the y-axis is in decibels using a reference pressure of 1psi. Hence when pulsed upstream the flow drops in pressure on encountering the blockage. There is a drop
of about 12dB at the 1Hz pulsing frequency. One of the important things to observe with the 1Hz pulsation cases is that the fundamental pulsation tone of 1 Hz has the highest amplitude. Subsequent harmonics can be observed at every 1Hz interval to about 20Hz. The fundamental tone amplitude will be studied in more detail throughout this thesis. The pressure recorded at the fundamental tone, for every opening and closing of the inlet, is studied to understand the pressure drops across the blockage.

![FFT and Pressure time Trace comparison](image)

**Figure 4-4: FFT and Pressure time Trace comparison**

### 4.3 Impact of Head

As mentioned in Section 3.4 the three main parameters that are changed in this study are head, blockage size and frequency. For the time being only 1Hz pulsations will be studied. First the impact of head is studied. 1” of H:0 corresponds to a pressure of 0.0361 psi. The pressure heads studied are 1”, 2”, 3” and 4” of H:0 which translates to 0.0361, 0.0722, 0.1083 and 0.1444 psi respectively.
4.3.1 Pressure Time Trace
Figure 4-5 shows a comparison of the pressure time traces at both port A and port F while simultaneously comparing them at 4in and 2in of head, respectively. At both ports, the 4in case has a much larger overshoot in the negative direction. The spike corresponds to the hole closing. This is due to higher flow rate associated with the 4in as against 2in. While comparing the time traces at the ports though it can be seen that Port F has less of an overshoot and has crests and troughs as compared to Port A. This is because downstream of the blockage the pressure reduces and the pulse dampens. The conservation of mass requires the flow rate upstream and downstream of the blockage to be equal. Basically the lower pressures seen at Port F prove that there indeed is a pressure drop across the blockage.

![P vs T](image1)

![P vs T](image2)

Figure 4-5: Impact of Head: (a) Port A (b) Port F

4.3.2 FFT
Figure 4-6 shows the FFT for the various heads. The case considered here is a blockage of 85% and the flow is being pulsed at 1Hz. As expected the greater the head, the greater the amplitude.
4.3.3 Pressure Drop

In Figure 4-6 the FFT is shown for just the 4in and 2in cases to get a clearer picture. Here the effect of head and blockage are studied together. The pressures in the pressure drop plots are calculated directly from FFT. The pressure in question is the fundamental tone pressure which is 1Hz in this case. This is calculated by taking the inverse logarithm of the amplitude corresponding to 1Hz. For the 85% blockage, the 4in and 2in heads are studied here. The pressure drop in the case of 85% shows that with a higher head there is a greater pressure drop. Figure 4-7 (a) shows the pressure drop across the blockage with respect to axial distance. The comparison is made between 2in and 4in of head and the plot shows how the pressure drops at 1 Hz from Port A to Port F across the blockage. It can be observed that the drop is more pronounced at 4in of head compared to the 2in of head.
Figure 4-7: Pressure Drop - (a) Axial Distance (b) Head

This pressure drop, $\Delta P$, is then plotted against head. Figure 4-7 (b) shows how the pressure drop increases linearly with head. The percentage pressure drop, $\%\Delta P$ can also be calculated using the following formula:

$$\%\Delta P = \frac{P_A - P_F}{P_A} \times 100$$

Where $P_A$ and $P_F$ are the values of pressures recorded at Ports A and F respectively. Figure 4-8 shows the effect of head for each blockage and how the $\%$ drop of pressure varies as a function of increasing head.
It is interesting to note that though there seems to be an increasing trend in the way absolute pressure drop rises with head, the percentage drop is more or less similar for all 4 heads. A comparison is made to theory. The theoretical points back up the experimental data, though the theoretical data is for steady state and not pulsating flow.

4.4 Impact of Blockage

4.4.1 Pressure Time trace

Figure 4-9 compares the impact of blockage in two separate plots. Here the pressure time traces are used for comparison at both Ports A and F. In the 96% case there is a drastic pressure loss across the blockage as can be seen in the plot. The pressure pulse downstream of the blockage, Port F, is almost a flat line at zero, indicating that there is hardly any pressure fluctuation between the systolic and diastolic phases. With respect to the human body, this is equivalent to a completely diseased artery. Also the wave form for the 96% lacks the large amplitude spikes due to lower velocity and hence lesser flow fluctuations.
In Figure 4-9 the 85% and 96% cases are shown accompanied by the corresponding flow visualization stills. The spike in the pressure pulse can be equated with the slight spike seen in the injected dye and this happens when the flow is blocked that is when the inlet is closed. In the 96% case the dye almost diffuses immediately once it passes through the blockage. This can already be seen in the 85% case which suggests that the 85% corresponds to diseased state.

Figure 4-10 shows the additional cases of 30% and 65% as well, accompanied by snapshots of the flow. The recording is at 3 inches of head. The dye doesn’t diffuse as much and the flow can be tracked downstream of the blockage.
4.4.2 FFT
Figure 4-11 shows the FFT for the 65%, 85% and 96% blockages. The 3in head case is considered and the spectra at Ports A and F are shown. The 65% blockage has the highest amplitude followed by the 85% and then the 96%. If the 1Hz peaks are observed, the 65% and 85% line up closer together downstream of the blockage compared to the 96% where the
spectrum amplitude is significantly lower. Logically speaking the spectra for each of the blockages upstream should have similar amplitude implying a constant upstream pressure irrespective of the blockage. At Port F, the lower amplitude for a higher blockage is understandable as the pressure drop is much greater when the flow is more constricted.

![FFT, Port A](image1)

![FFT, Port F](image2)

**Figure 4-11: Impact of blockage, FFT at Ports A and F**

4.4.3 Pressure Drop
The pressure drop is plotted against every head and every blockage in Figure 4-12. It is interesting to note that the increasing linear trend of pressure drop against head starts setting in with the 85% blockage. The 30% hardly shows any increase. The pressure drop across the blockage is almost negligible for the 30% blockage whereas from 65% onwards there is an appreciable pressure drop across the blockage as can be seen by the %ΔP values. In terms of pressure drop the percentage drops give greater insight into the state of the artery. The
percentage pressure is calculated as explained previously, \(\%\Delta P = (P_A - P_F)/P_A \times 100\). As the degree of stenosis increases there is a drastic fall in pressure compared to the upstream pressure and it is this relative fall that determines the healthy or diseased state of the artery. In terms of absolute pressure drop increasing head increases the pressure drop due to increasing flow rate.

![Graph showing pressure drop at different axial distances and heads](image)

**Figure 4-12: Pressure Drop, 4in Head**

### 4.5 Upstream vs Downstream Pulsing:

The facility has the capability of pulsating both upstream and downstream. The original setup had just the downstream capability and on the new improved set up a rotating wheel has been added in the upstream tank. The design of the rotating wheel is identical to the one downstream so a comparison can be made while studying the effect of head, frequency and blockage. The schematic is seen in Figure 4-13 In the case of upstream pulsing the wave and flow directions are same, whereas in the case of downstream pulsing the wave and flow are in opposite directions.
Figure 4-14: Schematic

4.5.1 Pressure Time Trace

Figure 4-14 shows the pressure time traces of a particular case: 3in head, 85% blockage and 1Hz pulsations. The solid lines denote upstream pulsing and the dotted ones denote downstream pulsing. The zero pressure line is the steady flow condition when both inlet and outlet are open as the 4in head drives the flow in this case. Looking at the pressure curves we observe that there are curves both on the positive and negative side. The positive pressure spike corresponds to the downstream pulsing. When pulsed downstream the outlet is periodically opening and closing due to the rotating wheel. At that instant when the outlet is blocked the flow is blocked which implies a reduction of the velocity to zero. Velocity reduction implies a rise in pressure in accordance with the Bernoulli equation.

On the other hand when pulsed upstream there is a closing of the inlet which is basically cutting off the effect of the 4in head which causes the pressure curve to dive in the negative direction. Also with the way the facility is set up there exists a small head at the outlet which can cause backflow when the inlet is closed. These factors contribute to the negative pressure observed in
the figure. In the real human set up this back flow is stopped by the closing of the Aortic semilunar valve which then prevents the back flow of the blood from the Aorta into the ventricle.

**Figure 4-14: Pressure vs Time**

In terms of flow visualization, the snapshots do not show a discernible difference between the upstream and downstream pulsing cases. Figure 4-15 shows the flow visualization for 85% blockage.
4.5.2 FFT

A comparison between the FFT spectra for the Upstream and downstream pulsing cases is shown in Figure 4-16. Pressure is measured at both the far upstream and downstream ports in both cases. The case presented here is 4in head, 96% blockage and 1Hz pulsation frequency.

![Figure 4-16: FFT Upstream vs Downstream Pulsing](image)
On the left is when the facility is pulsed upstream and on the right is when the facility is pulsed downstream. The most obvious observation is that the plots have flipped. When pulsed upstream the fundamental tone and the subsequent harmonics recorded are higher in amplitude at Port A rather than F. In the Downstream pulsing case, it is vice versa where higher pressures are recorded at Port F the downstream tap. As has already been seen in section 1.5 in Adewumi’s study [14] the location of pulsation and hence the wave propagation direction decides whether there is a pressure rise or a pressure drop across the blockage.

4.5.3 Pressure Drop
Figure 4-17 further quantifies this pressure drop vs rise for the fundamental tone. As can be seen in the upstream pulsing case there is a drop in the fundamental tonal pressure across the blockage and the downstream pulsing case there is a rise across the blockage. This pressure change is shown as ΔP.

**Figure 4-17: Pressure vs Axial Distance**
In Figure 4-18 $\Delta P$ is plotted against the head. This $\Delta P$ as mentioned earlier is the difference between the fundamental tone amplitudes recorded at Ports A and F. For both the upstream and downstream cases there is a linear increase in pressure drop/rise with increasing head. In both figures it can be seen that the 85% blockage case shows the trend developing and in the 96% case the trend is fully developed wherein the pressure change increases with increasing head.

![Upstream Pulsing](image1)

![Downstream Pulsing](image2)

Figure 4-18: Upstream vs Downstream Pulsing

### 4.6 Wheel Design: Duty Cycle

As explained earlier, a spinning motor attached to the rotating wheel is the component that causes the pulsation in the flow. The spacing of the holes and the rate at which they open and close the inlet are the factors that determine the pulse shape. Along with this, the spacing of the holes is important since this determines the duration of systole and diastole and hence the systole/diastole ratios (S/D). Figure 4-19 shows the two wheels that were designed and manufactured in addition to the existing wheel with 4 equally spaced holes. With the original
wheel the S/D ratio is 1.31. Typically in the heart the S/D ratio is less than 1. For the 1Hz case, V. Gemignani et al [6] suggest that a S/D of 0.5-0-6 is ideal, so a wheel with S/D= 0.523 was designed. This particular wheel has two holes diametrically opposite to each other with a diameter of 1.375” each. For the 2 Hz case the wheel has 3 equally spaced holes of diameter 1.125”, which leads to an S/D of 0.848. Now with the wheel comparison being taken into account the same FFT, pressure drops and trends are studied.

Figure 4-19: (a) Old Wheel (1Hz and 2Hz) (b) New Wheel 1Hz (c) New Wheel 2Hz

4.6.1 Pressure Time Trace
Figure 4-20 helps explain the impact of the duty cycle on the pressure time traces. The plot corresponding to S/D=1.31 has a longer flat line at zero pressure, that is the crest is present for a longer time than the trough. This shows that the hole is open for a longer time while spinning indicating a higher systole duration and hence a higher systole diastole ratio. The plot corresponding to S/D=0.523 meanwhile shows a longer diastole and this clearly shows a S/D ratio of less than 1. The difference in the nature of the curve after every second might have to do
with the slight difference in machining of each particular hole and the space between the wheel and the inlet which might cause a slight leakage even when the wheel covers the inlet.

Figure 4-20: Duty Cycle Comparison

Figure 4-21 (a) shows that at a heart rate of 60 bpm the S/D ratio needs to be about 0.5 to 0.6. Hence a 2 hole wheel was designed to obtain a S/D=0.523. As discussed in section 2.4 the systole diastole ratio rises as the heart rate increases. Later in section 4.7 under the impact of pulsing frequency, the 2Hz case is discussed in more detail.
Figure 4-21: (a) S/D ratio vs Heart Rate; (b) S/D=0.523, 1Hz New Wheel

4.6.2 FFT

Figure 4-22: FFT (a) S/D=1.31 and (b) S/D=0.523

Figure 4-22 (a) shows the FFT spectra of the old 4 hole wheel with an S/D=1.31 and (b) shows the new wheel for the 1Hz case with an S/D=0.523. For both wheel designs the fundamental tone amplitude is very similar. The case considered here is the 96%, 4in head.
Figure 4-23: FFT (a) Old Wheel-S/D=1.31 (b) New Wheel S/D= 0.523

Figure 4-23 shows a comparison between the spectra of the two wheels with respect to the changing blockage. The behavior of the wheel with S/D= 0.523 is along expected lines as the fundamental amplitude at 1Hz are almost same at Port A. These measurements have been taken at the upstream port, Port A. The case considered here is the 3in, 1Hz pulsation.

4.6.3 Pressure Drop

Finally a comparison is made between the pressure drops for both S/D ratios. The upper half of Figure 4-24 shows pressure drops against axial distance for three blockages. For the 96% blockage, the percentage pressure drop is almost similar for both S/D ratios. Irrespective of the systole/diastole ratio the 96% case is almost certainly diseased. For the 65% and 85% cases, the %ΔP is 10 to 15% lower when the S/D ratio is closer to the S/D for the human heart i.e. S/D=0.523. Also the upstream pressures recorded are almost equal for the S/D=0.523 case. This is believed to be to the longer times of no flow which allow the flow to fully come to rest. The downstream pressures show difference in values depending on the size of the blockage lower pressures for larger constrictions. This is in line with expected behavior of pressure drops across a stenosis. The pressure drop against head
shows similar trends in both cases as seen in the lower part of Figure 4-24. In the lower panel of Figure 4-24 an error bar analysis is done. Here repeatability as well as the slope were considered by using a best fit approach. An error of about 13% is seen in the case of the 3in 85% case. Figure 4-25 further emphasizes the comparison of pressure drop between the two different S/D ratios. The pressure drops for every head is lower for the new wheel (S/D=0.523). The biggest difference between S/D ratios is for the 85% blockage. This is a diseased case but less diseased compared to 96%. The 85% case does emphasize that a S/D ratio lower than 1 is more efficient through blockages. When comparing multiple cases (particularly heads), the percentage pressure drop becomes a better metric. From Figure 4-25 it is evident that %ΔP doesn’t change too much with head especially in case (b). However, the increase in %ΔP with increased blockage size is still apparent.
Figure 4-24: Pressure Drops: 1Hz Duty Cycle, 4in Head
Figure 4-25: Effect of changed S/D for each blockage

Figure 4-26: Percentage Pressure Drop S/D based on different S/D ratios[6]
4.7 Impact of Pulsing Frequency
So far the impact of head, blockage and duty cycle have been studied keeping the pulsing frequency constant at 1Hz. This corresponds to studying the body at rest. In addition to changing the duty cycle at the same frequency, the experiment was conducted at a higher frequency of 2Hz. This typically corresponds to a person under stress or strenuous physical activity when the heart beats at a faster rate trying to pump a greater amount of blood to various parts of the body. More often than not though the effect of the stenosis is felt more severely when the human body is under stress. This section takes a look at how changing the pulsing frequency to 2Hz affects the spectrum and the pressure drop. Also in the human heart the S/D ratios change according to the requirements of the body. At a higher heart rate the time duration the heart ejects blood is slightly higher than when the body is at a state of rest. The relationship between pulsing frequency and S/D ratio will be explored together since this is a better model of the human heart.

4.7.1 Pressure Time Trace
Just as in the case of 1Hz a second wheel has been designed in the 2Hz case to better simulate the S/D ratio in the stressed state. In the new wheel the systole time has been reduced. Since the wheel has to spin faster for 2Hz and the S/D ratio is increased, it was necessary to have three pulsing cycles occur for one rotation of the wheel. Figure 4-27 shows the S/D ratio for 2Hz case and the corresponding wheel. The pressure time trace of the 2Hz case is observed in Figure 4-28.
The 1 Hz S/D ratio of 0.523 had two cycles for one spin. The spikes in the pressure are more accentuated for the S/D=0.848 case which is attributed to the wheel spinning faster. Here the 3in, 85% blockage is considered. The decrease in the duration of systole is seen form the pressure time trace. More importantly Figure 4-29 compares the two pulsing frequencies, 1Hz and 2Hz. Time has been normalized here to compare the two different pulsing frequencies. The two signals have many similarities, but the biggest difference is the larger S/D ratio for 2 Hz
Figure 4-27: (a) S/D ratio vs Heart Rate; (b) S/D=0.848, 2Hz New Wheel
Figure 4-28: Duty Cycle comparison for 2Hz
4.7.2 FFT

Figure 4-30 shows the FFT of the 2Hz case for two different S/D ratios. The S/D ratio of 0.848 was specifically designed to match the S/D ratio for the heart pumping at a higher bpm. Both cases have a 3in head and 85% blockage. The main similarity in the two cases is that the fundamental tone is at 2Hz and the subsequent harmonics are occurring every 2Hz and is clearly identified to about 20 Hz. The roll-off frequency and amplitude is also similar for both cases. One main difference is that the S/D ratio of 1.31 has more energy at the fundamental tone and at the odd frequencies (3 Hz, 5Hz, etc) the amplitude is lower.

Figure 4-31 shows a comparison between the FFT spectra between 1Hz and 2Hz. While making this comparison for both cases, 1Hz and 2Hz, the new wheel or the S/D ratio closer to the human body is considered, S/D=0.523 in the 1Hz case and S/D=0.48 in the 2Hz case respectively. The most obvious difference lies in the fundamental tone. The maximum amplitude is recorded at
2Hz for the 2Hz case. In both the cases though it is interesting to note that at Port A the fundamental tones recorded for all 3 blockages are almost identical. This is not seen in the case when the S/D ratio was higher that is when S/D=1.31 (Old Wheel).

Figure 4-30: FFT, 2Hz, Wheel Comparison, 3in Head
4.7.3 Pressure Drop

Next the pressure drop for the 2Hz case is looked at in Figure 4-32 for two S/D ratios. A similar comparison as in the 1Hz is made with plots of pressure drops vs axial distance in the 2Hz case.

The interesting point here to note is that in the case of the lower S/D ratio, the pressures upstream of the blockage (at Port A) are all the same. On the contrary, the higher S/D ratio case has the pressure decreasing with blockage size. An explanation of this could be that by
simulating the systole and diastole durations closer to the human body the flow is given enough time to stabilize as the inlet opens and closes. A similar conclusion was made for the 1 Hz case.

![Graph](image)

**Figure 4-32: Pressure Drop across blockage for 2Hz pulsing case: (a) S/D = 1.31 (b) S/D = 0.848**

Similarly the pressure drop vs the head is plotted in Figure 4-33. Similar trends can be seen for both cases which were also consistent for the 1 Hz case. Increasing the blockage will increase the pressure drop, and increasing the head will also increase the absolute pressure drop.
Discussions on the pressure drops and certain trends will be seen in the next chapter. Next a closer look is taken at the effect of the duty cycle for each blockage separately in the case of the 2Hz case just as in the 1Hz case. In Figure 4-34, 65%, 85% and 96% cases each the pressure drops vs head are looked at for the two different duty cycles. In all 3 cases it is seen that for the lower S/D ratio the pressure drops are lower for every head. One way of looking at this is that for the lower systole diastole ratio which is closer to how the human heart functions a high head or blood pressure is not required to overcome the pressure drop.
5 Discussion

5.1 Analysis of Pressure Drop
This chapter looks closely at percentage of pressure drop across a blockage and how these drops are affected by the duty cycle as well as the pulsation frequency. These factors are parametrically studied in terms of head and percent blockage.
In the previous chapter it has been observed that downstream pulsing causes a pressure rise across the blockage and upstream pulsing causes a pressure drop. In addition to this depending upon head, the extent of the blockage the pressure drop values are different. Also the frequency of pulsing was another factor introduced into the experiment. This too affects the pressure drops.

Figure 5-1: Pressure Drop vs Duty cycle
Figure 5-1 shows plots of pressure drops. In addition it also shows the percentage drops too. This comparison has been made for the 1Hz cases so these in effect are plots comparing duty cycles. The 4in head case has been shown here. For a clearer picture, Table 5-1 and Table 5-2 below list the percentage drops for each head and blockage in a tabular form.

<table>
<thead>
<tr>
<th>%ΔP</th>
<th>4in</th>
<th>3in</th>
<th>2in</th>
<th>1in</th>
</tr>
</thead>
<tbody>
<tr>
<td>65%</td>
<td>31</td>
<td>5</td>
<td>14</td>
<td>29</td>
</tr>
<tr>
<td>85%</td>
<td>47</td>
<td>35</td>
<td>29</td>
<td>34</td>
</tr>
<tr>
<td>96%</td>
<td>79</td>
<td>75</td>
<td>74</td>
<td>71</td>
</tr>
</tbody>
</table>

Table 5-1 Percentage Pressure Drop,1Hz, S/D=1.31

<table>
<thead>
<tr>
<th>%ΔP</th>
<th>4in</th>
<th>3in</th>
<th>2in</th>
<th>1in</th>
</tr>
</thead>
<tbody>
<tr>
<td>65%</td>
<td>16</td>
<td>20</td>
<td>17</td>
<td>14</td>
</tr>
<tr>
<td>85%</td>
<td>35</td>
<td>32</td>
<td>17</td>
<td>31</td>
</tr>
<tr>
<td>96%</td>
<td>75</td>
<td>71</td>
<td>69</td>
<td>61</td>
</tr>
</tbody>
</table>

Table 5-2: Percentage Pressure Drop,1Hz, S/D=0.523

The highlighted boxes are the case shown in the two plots. It is evident for both the 85% and 96% cases that when compare the two duty cycles, the corresponding values for pressure drop for the same head is less in the case of the lower S/D ratio. As an example for the 3in head, 85% blockage case from both tables, the value in the table with S/D=0.523 is lower. This suggests that with a S/D that is closer to the value associated with the human heart the percentage pressure drops are lower, suggesting that the head/pressure at which the heart is required to pump in order to overcome a blockage of a certain size is lower the closer to the ideal systole diastole ratio. In the 96% case, the values are pretty similar indicating that the 96% is a highly diseased case and fatal. The two plots in Figure 5-1 which show pressure drop vs head are quite similar for both
the duty cycles and they also reiterate the aforementioned fact about how the percentage pressure drops are less in the case of a lower S/D ratio. The plot of pressure drop vs head is more or less linear for each particular blockage. The slope of this line is lower in the case of the lower S/D ratio (0.523) rather than for the higher S/D ratio (1.31).

Figure 5-2: % Delta P vs Head and Normalized Area, 1Hz
Figure 5-2 shows a group of plots that further emphasize the plots shown in Figure 5-1. The two figures above are plots of percentage pressure drops against head for each blockage. Increasing head does not significantly change the percentage drops for each blockage. Looking at the plot that shows the percentage change vs normalized area, the two plots for each duty cycle line up quite well. Normalized area is the area of pipe constricted divided by unobstructed area of the pipe. The double headed arrow tries to show the difference between the two S/D ratios for a particular blockage. An important conclusion from this is that changing the duty cycle has more of an impact for the 65% rather than 96% blockage as can be seen from the points being closer for the 96% case and farther apart for the 65% case. If the size of blockage is greater, the impact of reducing the duty cycle closer to the human heart is less. For the 65% case, the duty cycle is even more important and the 85% case shows a similar albeit less trend. This begins to explain why two patients may respond differently to a similar blockage. Thus the blockage geometry along with heart rate and duty cycle will ultimately determine the pressure drop across a blockage. The significance of these percentage pressure drops comes to the fore when the two different frequencies are compared.

<table>
<thead>
<tr>
<th>%ΔP</th>
<th>4in</th>
<th>3in</th>
<th>2in</th>
<th>1in</th>
</tr>
</thead>
<tbody>
<tr>
<td>65%</td>
<td>16</td>
<td>20</td>
<td>17</td>
<td>14</td>
</tr>
<tr>
<td>85%</td>
<td>35</td>
<td>32</td>
<td>17</td>
<td>31</td>
</tr>
<tr>
<td>96%</td>
<td>75</td>
<td>71</td>
<td>69</td>
<td>61</td>
</tr>
</tbody>
</table>

Table 5-3: Percentage Pressure Drops 1Hz

<table>
<thead>
<tr>
<th>%ΔP</th>
<th>4in</th>
<th>3in</th>
<th>2in</th>
<th>1in</th>
</tr>
</thead>
<tbody>
<tr>
<td>65%</td>
<td>23</td>
<td>17</td>
<td>14</td>
<td>14</td>
</tr>
<tr>
<td>85%</td>
<td>22</td>
<td>25</td>
<td>10</td>
<td>26</td>
</tr>
<tr>
<td>96%</td>
<td>59</td>
<td>53</td>
<td>55</td>
<td>55</td>
</tr>
</tbody>
</table>

Table 5-4: Percentage Pressure Drops 2Hz
Table 5-3 and Table 5-4 show the %\(\Delta P\) for the 1Hz and 2Hz cases. The comparison is made between the S/D ratios which are most common for each heart rate, that is S/D = 0.523 for the 1Hz case and S/D = 0.848 for the 2Hz case. Figure 5-3 shows percentage pressure drops for the 1Hz and 2Hz cases.

It can be observed that the pressure drops for the same corresponding cases are quite different. In the 2Hz case for the 65% and 85% blockages the percentage drops are much lower with more than a 50% lessening as compared to the 1Hz case. In the 96% case the lower percentage pressure drop in the case of 2Hz case is evident. This observation tends to suggest that keeping the extent of the blockage and head fixed, increasing the frequency of pulsation tends to lower the percentage drop of pressure. If we look at a particular blockage and change the heads it is seen that there isn’t too much of a change in values between different heads for both 1Hz and 2Hz cases. This suggests that the head doesn’t create too much of a difference in either case.
Figure 5-4 is similar to Figure 5-2 except that in this case a comparison between 1 Hz and 2 Hz is being made. Here too the percentage $\Delta P$ doesn’t vary too much with head. It is interesting to note though that for the 65% case percentage drop for 2 Hz is larger.

Figure 5-4: %Delta P vs Head and Normalized Area
5.2 Effect of Changing Setup:
During the process of conducting experiments various changes to the set up were made in order to improve the results. The changes made which did end up changing the FFT spectrum were, opening up the pressure measurement port in the test section and changing the size of the mount on which the pressure transducer is mounted. The two farthest ports on both the upstream and downstream sides were widened from about 0.2 mm to 5.5 mm in diameter. The next change was in the size of the brass mount which helps screw in the pressure transducer into the pressure port. By shortening the mounts the pressure transducer is brought closer to the flow which facilitates a better signal being picked up from the flow.

![Sensor improvement](image)

**Figure 5-5: Sensor improvement**

The objective here is to get the transducer as flush mounted as possible to the flow. Subsequently further changes were made in terms of taking readings with the test section turned 90deg from the original orientation. This section will take you through the results obtained from the various changes made in the experimental setup. Figure 5-5 shows how opening up the port further and subsequently bringing the transducer closer to the flow improves measurement. With mount 2 we
are practically right over the flow while taking measurements. The window in the figure highlights frequencies below 20 Hz. It is observed that the higher frequencies are still impacted by sensor mounting and with the current pressure transducer it is dangerous to draw any conclusions from the higher frequency spectra.

5.3 50Hz peak

Now that the FFT’s and the pressure time traces along with the pressure rise and drop trends have been observed it is time to take a closer look at the FFTs. So far the only the fundamental tone has been looked into for the pulsating flow. When the FFT is observed for slightly higher frequencies, as mentioned earlier by the 25Hz mark a roll off is observed. There is then a slight rise in the 40-60 Hz region. At 50 and 100 Hz there are slight peaks observed especially in the 96% case. The first plausible reason explored was that of mounting. On rotating the test section 90 degrees i.e. basically rotating the pressure transducer by 90 degrees from its initial position which was right on top of the flow it is observed that those peaks almost vanish and are not as prominent.
It is observed that even before any changes were made to the set up in terms of opening up the pressure port or changing the mounting using a shorter brass fitting this particular peak was observed. Figure 5-6 shows a comparison, the spectrum on the left being the current configuration and the one on the right the older configuration with the smaller hole for the pressure port and longer brass fitting. Though the markedly observable difference is the amplitude of fluctuation and the slight increase in frequency where the roll off begins, it is indeed seen that the 50 Hz bump can be observed for both as well as a sharp peak at 100 Hz. It is more pronounced for the 96% blockage case.

The other immediate comparison that was made was the upstream vs downstream pulsing. It was interesting to check whether the direction of pulsing was in anyway affecting the occurrence of these peaks at 50 Hz and 100Hz. Again it is observed that both upstream and downstream pulsing have the peaks. Figure 5-7 illustrates this.
Figure 5-7: (a) upstream vs (b) downstream pulsing

It is interesting to note in both the above cases discussed that these peaks become more prominent as the constriction is larger.

Similarly a comparison was done for the different wheel designs. In Figure 5-8 on the left is the wheel with a S/D = 1.31 and on the right is the wheel the new 2 hole wheel with a S/D= 0.523. There is nothing to suggest that change in duty cycle shifted or got rid of these peaks. In addition to all these cases, the experiment was tweaked slightly. The test section basically comprises of two halves as described earlier in the thesis with the blockage inserted in the middle. The test section is set up such that the pressure taps are in the upright position. Now the downstream test section is rotated 90 degrees such that the pressure taps are facing the experimentalist. This was done to make sure that air bubbles if any are eliminated completely when the pressure transducer is fitted into Port F. The results obtained when this was done are presented in Figure 5-9.
Figure 5-8: (a) S/D=1.31 vs (b) S/D=0.523

Figure 5-9: Rotated Configuration

In Figure 5-7, Figure 5-8 and Figure 5-9 the heads tested were 4in. To make sure that the head did not have any impact an FFT comparison for the various heads was also done and it turned out that it is just the amplitude that changes with head change but as such a similar spectrum is
obtained including the peaks. This seems to suggest that the pressure head does not affect the nature of the flow itself but just changes affects properties such as flow rates and velocities that are discussed later.

5.4 Flow Rates
Literature shows that wall pressure fluctuations have been studied both in rigid and elastic pipes. A.O. Borisyuk [26] in his study on wall pressure had found that the value of the rms pressure tended to increase immediately downstream of the narrowing or constriction and reached a maxima just at the point of jet reattachment. When the power spectrum was observed it revealed that there is a maxima in the low frequency region and have been associated with eddies in the flow separation and reattachment regions. Figure 5-10 below shows the how the blockage has been set up inside the tube and how the reduction in diameter causes a turbulence downstream of the narrowing resulting in a flow separated and a reattachment region. After this the redevelops to the same as the upstream flow structure. I- in the figure shows the flow separation region II- shows the reattachment region and III- shows the region where the flow redevelops and stabilizes. Our current set up is very similar to this sort of configuration in terms of the type of blockage insertion the abrupt narrowing of the flow and hence the flow characteristics.

The mean axial flow velocity was determined by finding out the flow rate by collecting for a fixed amount of time the water from the outlet. Then the velocity was determined using the relation (write relation). In the current experiment the effect of head and blockage is studied together and each blockage in tandem with the head results in a different volume of fluid flowing and hence different flow rates and velocities.
Table 5-5 below shows the flow rate \('Q'\). With the help of these flow rates the mean velocities can be computed using the relation,

\[
U = \frac{Q}{T \pi D^2 / 4}
\]

(4.1)

where \('U'\) is the mean velocity in the test section, \(T\) the time period of data acquisition and \('D'\) the diameter of the test section. The mean velocity through the constriction is computed using the conservation of mass condition

\[
u = U \left(\frac{D}{d}\right)^2
\]

(4.2)

Here \('u'\) is the velocity through the constriction and \('d'\) is the diameter of the blockage. As per the experiment conducted by the values of \('U'\)<0.39m/s were only considered since the \(Re\) range in the larger arteries for in the body is typically below 7000 especially the ascending aorta.
<table>
<thead>
<tr>
<th>Flow Rates(l/s)</th>
<th>65%</th>
<th>85%</th>
<th>96%</th>
</tr>
</thead>
<tbody>
<tr>
<td>4in</td>
<td>0.14+/-0.018</td>
<td>0.075+/-0.003</td>
<td>0.02+/-0.0003</td>
</tr>
<tr>
<td>3in</td>
<td>0.12+/-0.007</td>
<td>0.068+/-0.002</td>
<td>-----</td>
</tr>
<tr>
<td>2in</td>
<td>0.10+/-0.005</td>
<td>0.05+/-0.0006</td>
<td>-----</td>
</tr>
<tr>
<td>1in</td>
<td>0.098+/-0.003</td>
<td>0.04+/-0.0005</td>
<td>-----</td>
</tr>
</tbody>
</table>

Table 5-5: Flow rates[6]

In terms of the heart and arterial stenosis, the flow rates in relation with pressure drops and radii have been studied. The concept of critical stenosis is looked into as well. Figure 5-11 below shows flow as both a function of radius and its relation to delta P or drop in pressure.
According to Poiseuille’s equation for laminar flow flow is proportional to ΔP and also to the 4\textsuperscript{th} power of radius. With the presence of turbulence though the graph changes as can be observed in the figure above. The equation

\[ F \propto \frac{\Delta P r^4}{\eta L} \]

Here ‘F’ signifies flow, ΔP the pressure drop, ‘r’ the radius,’\eta’ viscosity and ‘L’ the length. One of the conclusions that can be drawn from this is that a 50% decrease in radius corresponds to a 16 fold decrease in flow.

As mentioned earlier flow visualization was carried out in order to be able to observe the nature of the flow for various blockages.
Figure 5-12: Flow visualization 3in 0% blockage

Figure 5-12 shows an unobstructed test section and one can see the formation of a small vortex just downstream of the blockage. Figure 5-13 meanwhile shows a 96% blockage and one can see how due to the heavy obstruction the 96% acts like a nozzle and the dye diffuses into the downstream side.

Figure 5-13: Flow visualization 3in 96% blockage
5.5 Conclusions
The current study is experimental and was successful in quantifying pressure drops for the various blockages pressure heads and pulsation frequencies. The FFT spectra and the pressure time traces were looked at and flow rates were measured.

Firstly a great many changes were made to the previous pulsing facility in adding more parameters and additional diagnostics. The first and most significant change made was the addition of an upstream pulsing capability. Some of the major findings due to various design and diagnostic changes are listed below.

5.5.1 Upstream Pulsing
This change made sure that the facility resembles the human body in terms of having both the pulse and the flow in the same direction, just like in the human aorta. A comparative study between upstream and downstream pulsing was done and it was observed that on pulsing upstream pressure dropped across the blockage which is as expected whereas in the downstream pulsing case there was a pressure rise across the blockage.

5.5.2 Blockage
In the current study an additional blockage was fabricated and studied. Previously a 65% medium blockage was followed by a 96% blockage which corresponds to a completely diseased state. An 85% blockage (area reduction) was fabricated to understand and try and identify a critical extent of blockage. According to Conley et al [27] diameter narrowing of 75% (94% area) was initially defined as significant. But further studies later concluded that 50% diameter narrowing (75% area) was significantly diseased. Considering that there is little information about degree of stenosis on clinical outcome it was decided that a diameter narrowing of 60% (85% area reduction) should be studied. Pressure drop trends showed that the behavior of the
85% blockage was in between that of the 65% and 96%. The flow visualization however showed more similarities with the 96% case.

5.5.2.1 Eccentricity and Asymmetry of Blockage

It is important to note the effect of stenosis geometry on stenosis haemodynamics. S.R. Dodds [28] studied the effect of asymmetric stenoses on arterial haemodynamics. Experiments were conducted using a concentric circular stenosis, offset circular stenosis and an eccentric or non circular lumen offset from the central axis. The three different types can be seen in Figure 5-14.

It was concluded finally from the experiment that there is flow dependent resistance irrespective of the geometry. The effect of asymmetry was seen on the hameodynamics only if the stenosis severity was expressed as a diameter reduction and not as an area reduction.

The resistance to flow rate relationship was found to be linear for all geometries where resistance \( R = \frac{\Delta P}{Q} \) such that \( R = K_1 + K_2 Q \). Figure 5-14 shows the different stenosis geometries. In conclusion rigid non circular stenosis have the same hydraulic effect as circular stenosis of same area reduction. The same can’t be said if the severity is expressed as a diameter reduction.
Flachskampf et al [29] studied the effect of orifice geometry and flow rate on effective valve area. An in vitro model was used in the study to examine the discharge coefficient by varying the 1) flow rate 2) valve area 3) inlet geometry 4) orifice eccentricity. Different circular orifices were used with areas varying from 0.3 cm² to 2.5 cm². Three different elliptic orifices were used with eccentricities 2:1 3:1 and 5:1. Figure 5-15 below shows the combined effect of varying area and eccentricity. The equation slopes of the lines were obtained using multiple regression analysis. Here the discharge coefficient (DC) is the independent variable and Area and eccentricity (ECC) the independent variables. To better understand and emphasize the effect of these two independent variables the Table 5-6 shows the individual effect of area and eccentricity on the coefficient of discharge. Since stenosis geometries are known for highly variable geometry coefficient of discharge is expected to vary. In conclusion it was observed that the on
changing area from 0.3 to 2.5 $cm^2$ the Cd increased by a mean of 8.9%. On varying the orifice from an eccentricity of 5:1 to circular, Cd increased by 5.9%. Also it was observed that having a nozzle like inlet to the blockage implied greater Cd as against an abrupt blockage.

Figure 5-15: Combined Effect of Area and Eccentricity[29]
Table 1. Discharge Coefficients in Valve Orifices of Different Sizes and Shapes

<table>
<thead>
<tr>
<th>Area (cm²)</th>
<th>1:1 (circular)</th>
<th>1:1 With Nozzle</th>
<th>2:1</th>
<th>3:1</th>
<th>5:1</th>
</tr>
</thead>
<tbody>
<tr>
<td>0.3</td>
<td>0.72</td>
<td>0.81</td>
<td>0.71</td>
<td>0.70</td>
<td>0.68</td>
</tr>
<tr>
<td>0.5</td>
<td>0.73</td>
<td>0.83</td>
<td>0.72</td>
<td>0.71</td>
<td>0.70</td>
</tr>
<tr>
<td>1.0</td>
<td>0.75</td>
<td>0.85</td>
<td>0.74</td>
<td>0.73</td>
<td>0.71</td>
</tr>
<tr>
<td>1.5</td>
<td>0.77</td>
<td>0.88</td>
<td>0.76</td>
<td>0.74</td>
<td>0.73</td>
</tr>
<tr>
<td>2.0</td>
<td>0.79</td>
<td>0.90</td>
<td>0.77</td>
<td>0.75</td>
<td>0.73</td>
</tr>
<tr>
<td>2.5</td>
<td>0.82</td>
<td>0.93</td>
<td>0.77</td>
<td>0.74</td>
<td></td>
</tr>
</tbody>
</table>

Table 5-6: Effect of Area and Eccentricity on Co-Efficient of Discharge[29]

5.5.3 S/D ratio and Pulsing Frequency
Another significant change made was the design of the spinning wheel and hence the systole/diastole ratio. It was observed that by simulating S/D ratios closer to the actual human heart, the pressure drops percentages significantly reduced. More accurate conclusions can be drawn then as to the effect of a stenosis when the S/D ratio is accurate. Previously the S/D ratio was much higher (valve open for longer) and greater pressure drops were observed. With the more accurate S/D ratios it can be observed that the heart in reality does not need to work as hard to overcome the same extent of blockage. Also the 2Hz case suggests that when there is a severity of blockage by spinning the wheel faster at the correct S/D ratio percentage pressure drop decreases.
5.6  Future Work
This current facility successfully studied the effect of varying size of blockages, heads, pulsing frequency and duty cycle. There are many improvements and future work that needs to be done to understand stenosed artery better and to be able to relate the conclusions to actual clinical outcomes. Changes based on design specifics and diagnostics are seen here.

Firstly an additional blockage(s) needs to be fabricated. A 75% area reduction would be an ideal stenosis to study [27]. Additionally if the blockages can be made asymmetric or eccentric as discussed in 5.5.2.1 the blocked artery can be better modeled as plaque build up in an artery is rarely symmetric.

Secondly the effect of wall compliance needs to be studied in the future.

5.6.1  Arterial Wall Compliance
A study on the arterial wall compliance on the pressure drop across coronary artery stenoses was conducted by Banerjee et al [30]. The study was conducted for a wide range of stenosis severities. A comparison was made between a rigid artery, a compliant artery with calcified plaque and a compliant artery with smooth muscle proliferation under hyperemic flow conditions. It was observed that for a significant stenosis severity the pressure drop was lower by 27.7% and 37.6% for the calcified plaque and the smooth muscle cell proliferation case, respectively. The degrees of stenosis studied were 70% (moderate), 80% (intermediate) and 90% (severe) area stenoses. Additionally a decrease in ΔP by 5.7%, 9.3% and 27.7% was observed for the calcified plaque compliant model when compared to the rigid artery model. The significant difference in pressure drops could cause an overestimation and misinterpretation of stenosis severity.
Specific to this facility the design of the rotating wheel can be further changed. This facility simulates an accurate S/D ratio but further work needs to be done in order to achieve the exact pulse shape. This can be done by changing the geometry and spacing of the holes.

In terms of diagnostics a number of things can be improved. Firstly the mounting of the transducer and the number of transducers. Flush mounting it over the flow allows for a more accurate pressure recording. Also, by having multiple transducers the pressure profile can also be traced and phenomena such as pressure recovery can also be observed in addition to just the pressure drop.

Secondly the pressure ports need to be ideally slightly further upstream and downstream of the stenosis to account for the pressure recovery. Properties such as density and viscosity can be changed by adding cornstarch to the water. Improvements to the test section would also improve the results and compare more favorably to previous results obtained from studies. In actual fact the arteries and arterioles in the body are of much lower diameter of about. The current study had a test section with internal diameter 1in which is similar to that of the aorta. Along with this if the wall can be made compliant the test section would then simulate better the contractions and dilations of a blood vessel better

To make the work more comprehensive a comparison must be made between experimental and computational studies done earlier. Ghalichi [31] performed one such computational study in which low Re-no turbulence was compared to a laminar model. This predicts that turbulence can develop even at a blockage of 25%. Also in this study non concentric blockages have been used which is a more accurate simulation of a stenosis though only steady flow was studied. With the current experimental set up trying to recreate the variables in a computational environment
would help us understand and validate the results obtained far better. The boundary condition at the inlet needs to be equivalent to the pulsation pattern produced by the spinning wheel. In order to improve the experimental results better and more sensitive pressure transducers would be required. In addition if multiple pressure transducers can be flush mounted the pressure recovery pattern can also then be understood. Also with better flow visualization methods the turbulence could be observed and compared to experimental results such as those obtained by Ghalichi. Figure 5-16 shows recirculation zones for various blockages.

![Figure 5-16: Recirculation Zones for various blockages](image_url)

The computational challenge for this current facility would be to simulate the exact boundary conditions for the pulsatile flow and be able to introduce the exact pressure conditions.
6 References

7 Appendix

7.1 Codes [1]

7.1.1 FFT Code

clc; clear;
drive = 'F:\DATA';
save_loc = [drive 'FFT\'];
%open_loc = [drive 'Scott Data\Pressure Measurements\'];
open_loc = [drive '\New Wheel\'];
line_color = ['b' 'b' 'k' 'k'];
fs = 4096;
block = 10; % number of cycles to study
N = 16384;
%frelq=fs/N:fs/N:
freq=0:fs/N:fs-fs/N;
% Sets up file name to load each specific case
count = 0;
for blockage = 96
    count = count + 1;
    %file_name = ['Far_Fore_No_Aft_Port_ ' num2str(blockage)
    '%_1_Hz_Short_Tube']
    file_name = ['No_Fore_Far_Aft_Port_ ' num2str(blockage)
    '%_1_Hz_Short_Tube']
    sig = load([open_loc file_name '.txt']);
    sig = sig * 0.4856 * 0.036130068; % Converts volts to in H2O and in H2O to psi then psi
    sigavg = sig(1);
    sigstdev = sig(2);
    points = length(sig);
    sig2 = sig(3:points);
    delt=1/fs; %time per cycle
    T = delt*N; % total time run
    w=hanning(N);
    finish_point = 0;
    for bk = 1:block
        start_point = 1 + finish_point;
        finish_point = start_point + N - 1;
        sig_analyze = sig2(start_point:finish_point);
        sig_analyze = chanvals(start_point:finish_point);
        [bb, aa] = cheby1(5, 0.5, 20/(fs/2), 'low');
        %sig_analyze_temp = filtfilt(bb,aa,sig_analyze);
        sig_w=sig_analyze'.*w;
        y_fft=fft(sig_w,N);
        PHI_yy=2*abs(y_fft)/N;
        PHI_yy_amp(:,bk) = 20*log10(PHI_yy);
    end
    PHI_yy_amp_avg = mean(PHI_yy_amp,2);
a=10^((PHI_yy_amp_avg(5))/20)
b=10^((PHI_yy_amp_avg(17))/20);
D=0.8*0.0254;
v=0.032;
St= freq*D/v;
figure(2)
semilogx(freq,PHI_yy_amp_avg,'Color',line_color(count));
% Pressure Time trace
% Converts voltages to pressure and then plots against time

drive = 'F:\DATA\New Wheel 1\Horizontal\Far_Fore_No_Aft_Port_' % Accesses file from this location
save_loc = 'F:\Converted Pressure\'
file_save_loc = 'F:\Time Trace\'

% Initial Parameters
head = '1';
frequency = '1';
fs = 4096;
fr = frequency;
blockage = '65%';
h = 1; % head initializer
f = 1; % frequency initializer
b = 1; % blockage initializer

block = 25; % number of cycles to study
N = 500000/block;

file_name = [blockage '_' head '_' in_' frequency '_Hz_Short_tube.txt'];
% Since drive and file_name are separate strings, must but [] around

sig = load([drive file_name]);
% if need to convert to Pa, conversion is 6895 Pa per psi
sig = sig * .4856 * .036130068 ; % Converts volts to in H2O and in H2O to psi then psi

sigavg = sig(1);
sigstdev = sig(2);
% Next two lines remove average and st deviation from matrix
points = length(sig);
sig = sig(3:points);
[bb, aa] = cheby1(5,0.5,8/(fs/2),'low');
sig_analyze_temp = filtfilt(bb,aa,sig);
delt = 1/fs; % time per cycle
T = delt*N; % total time run
% sets matrix starting at 0, increasing by change in time for the
% total number of points in the signal matrix
time = delt:delt:(points-2)*delt;

save([save_loc blockage '_ Blockage_' head '_' in_' freqs '_Hz_Pulsations_Pressure'], 'time', 'sig_analyze_temp');
load(['save_loc blockage '_Blockage_ ' head ' in ' freqs '_Hz_Pulsations_Pressure'],[time,'sig_analyze_temp']);

    title_name = 'P vs Time for ';
figure(1)
hold on;
plot((time-0.075),{sig_analyze_temp});
title(['[title_name blockage ' Blockage_ ' head ' in Head ' freqs ' Hz Pulsations'],['FontSize',18]);
axis([0 max(time) -.20 .20]);
xlabel('\bfTime (s)',['FontSize',18])
ylabel('\bfPressure(Psi)',['FontSize',18])
%legend('Long Tube','Short Tube','Location','NorthEastOutside')
set(gca,'GridLineStyle',':',['FontSize',18]; grid on;
set(gcf,'color','w')
set(gca,'ytick',[-0.20 -0.15 -0.10 -0.05 0 0.05 0.1 0.15])

% Following Code Maximizes For Your Screen
%screen=get(0,'screensize');
%offset1=34;
%offset2=111;
%set(gcf,'units','pixels','Position',[1 offset1 screen(3) screen(4)-offset2])

% Saves File
saveas(gcf,['file_save_loc blockage ' ' head ' in ' freqs ' Hz'],'bmp');