DEPARTMENT OF MECHANICAL ENGINEERING

MECH3002 Individual Project

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Project Title:	**Development of in-house Tissue Engineering Bioreactor to test Heart Valves**
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I, Julia Elkouby, confirm that the work presented in this report is my own. Where information has been derived from other sources, I confirm that this has been indicated in the report.

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Abstract

Cardiovascular diseases (CVD) is one of the main causes of mortality and morbidity in the modern world with 1.6 million patients being admitted in the NHS in the UK alone (1). One of the most severe problems is aortic stenosis, which is the narrowing of the heart valve due the valve becoming calcified and unable to open and close properly.

To resolve this, patients can have an implant of either a biological heart valve or a mechanical one which both have their own benefits and disadvantages. Mechanical ones require medication of blood thinners which have many side-effects. The main drawback of biological ones is that the calcification of the heart valve will occur again and there is a high chance of reoperation (2,3). Due to the nature of this being a disease seen more often in older patients, a second operation is likely to be too physically demanding to the patients.

Researchers have been trying to observe what is happening to the heart valve so that they can investigate exactly what is affecting the calcification and what can be done to improve the expectancy of such a valve only lasting approximately 10 years. (2) In this project, therefore, an in-house bioreactor that mimics biomechanical conditions is designed and developed as an effective, customisable and affordable tool in heart valve research.

Based on the first, very simple version of prototype developed in the past, a second version is designed with the purpose of allowing researchers to see the valve through a viewing window clearly as well as mimicking the pressure on a heart valve between 80mmHg to 120mmHg. The flow of the water should also be in pulses allowing the flow rate to reach a peak of 209ml/s as in an aortic heart valve and then slow down. There should also be temperature control to allow the water to stay constant at the standard body temperature, 37^{o}C.

This prototype will be built and controlled using LabVIEW program. This will allow researchers with no programming experience to easily control the machine and test their heart valves as the interface would be easy to use and understand. This would allow Researchers with a non-technical background to use the machinery so that there can be more advances in this area.

The prototype of this project will also be constructed in a way that is easily dismantled and put back together to allow for easy modular changes to be made as the project advances so that more aspects can be monitored and controlled in its own time.

The performance of the bioreactor was then tested by calibrating and testing each individual control system to see if it mimicked the biomechanical parameters set at the start. Further testing and modifications are needed to mimic biochemical aspects of the heart in this machine.

Acknowledgments

This project was a great opportunity but also a challenge. There were many challenges along the way as well as many people whose support throughout was indispensable.

Firstly, I would like to thank UCL for their resources and easy support in allowing me to have access to lab spaces, professional workshop and all the software necessary to complete the project to a good standard. It was invaluable in helping to give me the space to explore the best solution for the project and the access was beyond the call of duty.

I would also like to thank my family and close friends without whose support I would not have been able to complete this project. In particular, my close friends at university who supported and were always ready to bounce off ideas. Also to my brothers and father who would always make me laugh when I was stressed and would always try to understand the problems I was facing. To my mum for helping me time manage and always having encouraging words of motivation as well as my boyfriend who was always there as a strong shoulder to lean on.

I would also like to thank Dr Ryo Torii for his unrelenting support every week while on the project and help in making the learning process a smooth and easy task with as little stress as possible. It made the project enjoyable and it was a pleasure to work so closely with him.

Contents

Nomenclature

Symbol	Meaning
CVD	Cardiovascular Diseases
AS	Aortic Stenosis
AV	Aortic Valve
SM	Stepper Motor
ρ	Density
L	Length
D & d	Diameter
P	Pressure
Q	Flow Rate
T	Temperature
V	Voltage
RPM	Rotations Per Minute

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1. Introduction

1.1 Background Information

Heart disease causes the most deaths worldwide. In the western world it is particularly prevalent with 5 million Americans being diagnosed with heart valve disease every single year (4). In the UK, 26% of males and 18% of females die prematurely of cardiovascular diseases (CVD) (5). Aortic Stenosis (AS) is one of the most common of these diseases.

AS is the narrowing of heart valves due to calcium depositing itself from the blood onto the heart valve tissue (6) caused by malformation or due to age. In aged patients' symptoms start to appear normally around 10 years after the calcium has started building up (6). The calcium deposits mean the tissue becomes much more rigid and causes the heart valve to not be able to fully open and close anymore shown in Figure 1. This means that the heart muscle cannot pump efficiently and leads to muscle fatigue and eventually heart failure (7). This is due to the increase in pressure on the heart valve as the same volume of blood is pushed through a smaller gap (1). This can also cause high blood pressure and can increase the chance of stroke and cardiac infarctions.

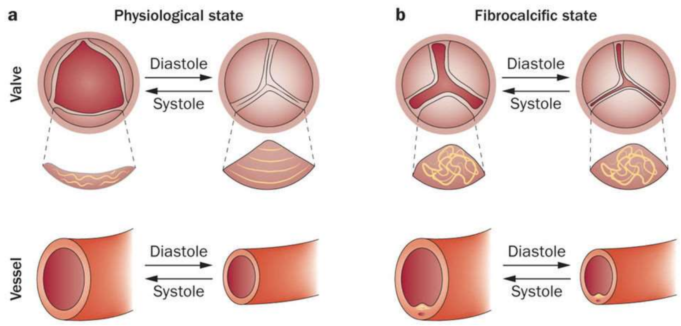

Figure 1: Figure showing the difference between a healthy heart valve and one that has undergone aortic stenosis (8)

To resolve this issue, the patients have to undergo a surgery to replace the Aortic Valve (AV). There are two types of AVR; Mechanical and Biological.

Mechanical valves are durable and last many years. However, due to the fact that they are made of materials such as carbon, the patients have to take anti-coagulant medication (2,9). This type of medication has many side effects, which include bleeding from other bodily functions as well as severe discomfort (10).

Biological heart valves do not require anticoagulant medication. However, the calcification occurs again, with a reoperation needed after an average of 6 years (2). This re-operation is a severe burden to the patient, also considering the age of these patients tends to be high, 22.2% of patients die within a year of reoperation (2,11).

Initial research shows that the main cause seems to be that collagen is present in biological heart valves. The collagen fibre degenerates due to the cyclic compressive and tensile stresses applied on the tissue. This means that calcium bonding sites are exposed, and the valve calcifies much more rapidly (12). To reduce this, the ideal solution would be to make a heart valve that is designed in such a way as to not undergo all these compressive stresses. There are ways to measure certain aspects in vivo such as through magnetic resonance imaging (MRI), but the accuracy is not very high, e.g. spatial resolution of MRI is typically 1mm per pixel (13). In order to realise a more in-detail observation of heart valve in operation, an in vitro bioreactor is effective, enabling to test different heart valves and see which ones last longer and decrease the risk of mortality after AV Replacement operations.

1.2 Literature Review

There have been many advances in the technology available to researchers to allow them to research different parts of the body tissues, including many around the cardiac one. For these types of research bioreactors were designed and built for purpose. One example is how bioreactors with pulsatile pumps have been made to create vascular grafts for replacement tissue (14,15,16). These are made in ways that mimic the conditions in heart vessels but are used to build up the tissue rather than test it to its limits. This is due to the concept that harvesting the tissue under the right conditions builds them in the most resistant way.

Many of these examples of bioreactors in research utilise peristaltic pumps already available on the market discussed in the competitor analysis; such as Hoerstrup Pulsatile Bioreactor (17), which utilises the Harvard pump (18) connected a silicone diaphragm which creates then the pulsatile flow of fluid to test the heart valves. This was one of the first bioreactors to use these techniques to test heart valves. However, procuring one of these pumps is not the cheapest and most effective solution and the fact that there are many attachments to different parts, and therefore large amounts of tubing, increases the chance of leaks and other malfunctions.

Although they mostly focus on vascular tissue grafts all the machines in these studies do mimic heart conditions to certain degrees and much can be learnt and used in the way they do it. In the multi-cue bioreactor (19,20) the pressure is increased by pinching the tube to adjust the pressure by decreasing the diameter of the tubing. There are also many more chemical aspects in that experiment as well as the mechanical ones which show how the two can work in conjunction with each other. The limitations of this are, however, that the study mostly focuses on the effect of the strain on the vascular tissue. One aspect that is useful is that all these machines are put together using screws to allow easy access to the grafts for microscopic analyses, which is useful when examining biological structures.

Another design instead used a piston pump to create the pulsatile flow for testing the left ventricle of the heart. This had many advantages and seems similar to the project that I will be undergoing. But as it is just built to measure the flow and pressure it does not provide clear visibility of the heart valve to the researcher (11). This is not vital in many experiments, but is a key aspect of being able to see the heart valve calcify in this particular case.

<table>
<tr><th colspan="2">Table of figures of Bioreactors from Literature</th></tr>
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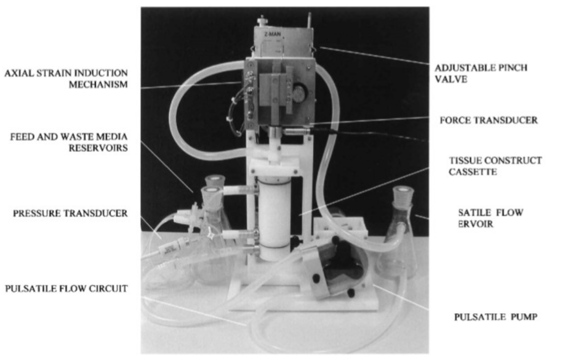

FIG. 1. Major components of the multi-cue bioreactor (MCB).

Figure 2:Diagram of Multi-cue Bioreactor (19)
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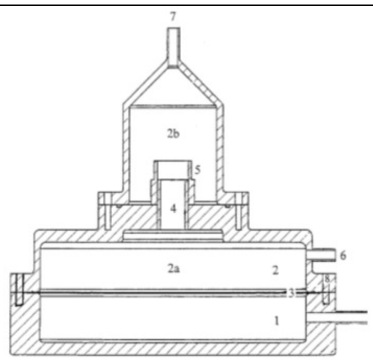

FIG. 2. Bioreactor (technical description).

Figure 3:Technical Diagram of Bioreactor (17)
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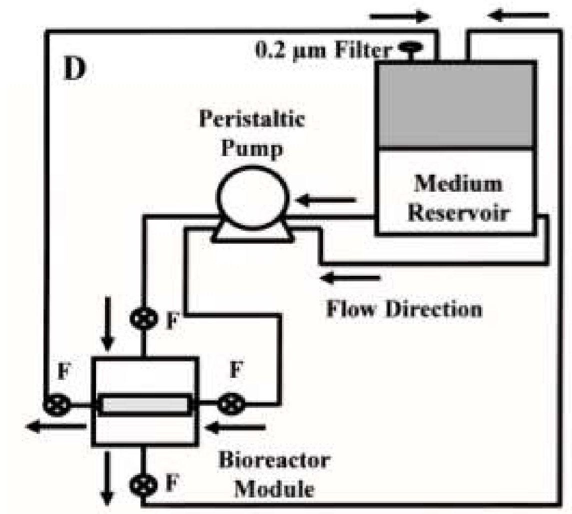

Figure 4:Diagram of Perfusion Bioreactor (20)
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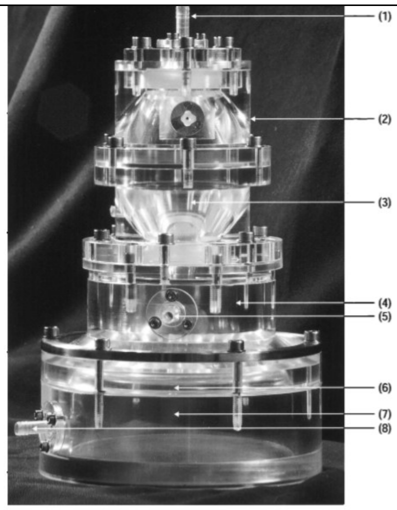

FIG. 1 Pulsatile bioreactor. (1) Cell culture outlet; (2) cell-seeding chamber; (3) vascular, polymeric scaffolds; (4) cell culture medium chamber; (5) cell culture inlet; (6) silicone diaphragm; (7) air chamber; (8) air inlet and air outlet.

Figure 5:Picture of Pulsatile Bioreactor (16)
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1.3 Competitor Analysis

Beyond the research bioreactors built in-house there are two companies who produce a pulsatile pump which mimics heart conditions in large quantities; Harvard Apparatus and Vivitro labs. These pumps both produce a pulsatile flow, however, they don't necessarily mimic the exact heart conditions. The Harvard Apparatus only produces a maximum of 100ml per stroke, which is significantly less than the 210ml per stroke needed for an aortic valve (18). These are also very expensive and cumbersome. Both are useful to a degree, but do not allow researchers flexibility in modulating it for research and the technical difficulty in understanding these machines also limits the research it can undertake.

Table 1: Table showing a comparison point between the Vivitro pump and the Harvard apparatus pump. Both leaders in this market (21,18,22).

	Harvard Apparatus	Vivitro Lab
	 *Figure 6:Picture of Harvard apparatus Pulsatile Pump (18)*	 *Figure 7:Picture of Vivitro Superpump which produces a pulsatile flow (22).*
Price (£)	£10,000	£21,140
Dimensions (cm)	50x21.2x33.7	53x92x45
Weight (kg)	14	18
Displacement Volume (ml/stroke)	0-100	0-180

1.4 Aims & Objectives

Aims

The main aim of the project is to design and build a bioreactor that simulates the haemodynamic conditions in the heart and aorta so researchers can test and observe the calcification of biological heart valves. The biomechanical factors critical to simulating blood flow through the heart valve which include, temperature, pressure and flow speed will be simulated.

Objectives

The software should;

- be easy to understand for non-engineers and easy to manipulate if needed
- show the flow speed and stepper motor RPM to allow a feed forward control system
- run the temperature control of the system

The Design should;

- be able to run for long periods of time
- allow a clear view of the heart valve throughout the experiment
- be easily dismantled for multiple tests and cleaning
- be structured in a way that is customisable for further research
- allow different pressure so that pulmonary and aortic heart valves can be tested
- only incorporate biocompatible materials
- not exceed the cost of £400

1.5 Design Requirements

After analyzing these objectives a final table of Design Requirements was created to make the process as easy as possible.

Table 2: Table of Requirements

Requirement Table	
Flow Rate	209ml/s in Pulsatile form
Temperature	37.5^{O}C
Pressure	80mmHg - 120mmHg

Pump Design	Durable for long-term use
Software Used	LabVIEW
LabVIEW interface	Easy to use
Viewing Window	Valve easily seen
Functionality of Machine	Easy to dismount and clean.
Cost	Under £400
Biocompatible Materials	Silicone, Acrylic

The final Bioreactor (Figure 8) shows the result of these requirements. In the production each design decision made will be detailed as well as how it relates to the final design. This would mimic the biomechanical heart conditions and allow researchers to see how the heart valve calcifies.

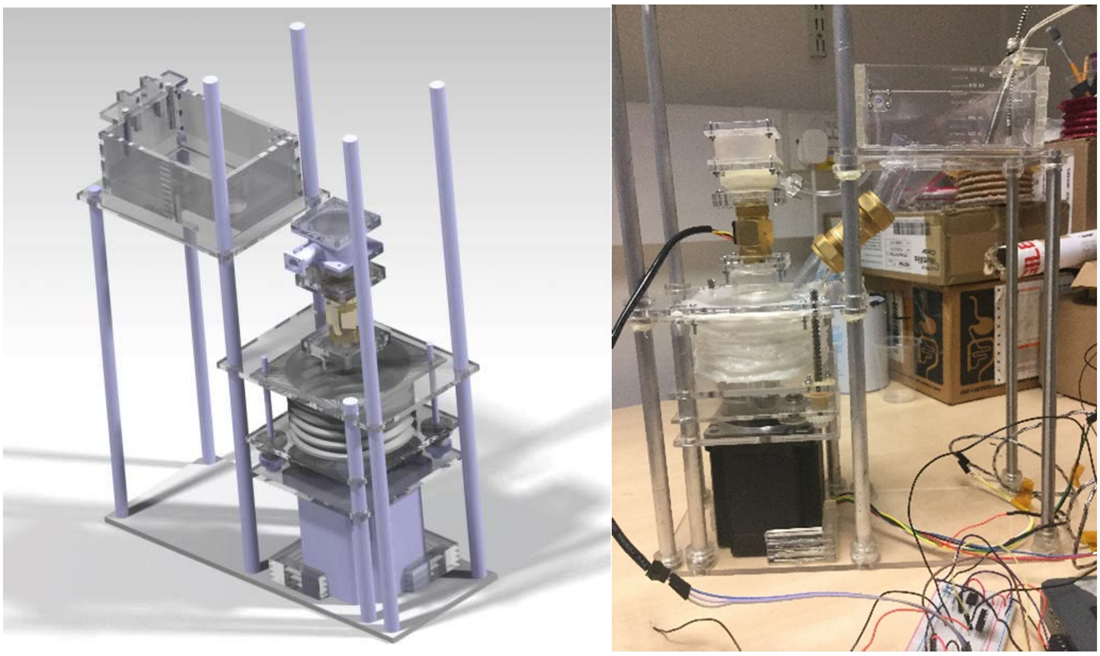

Figure 8: CAD Rendering and photo of Final Bioreactor Design

2. Design

The whole design requires many stages in designing and production. To make this more sequential, the system was divided into each control part that would be monitored or controlled. These are detailed in the diagram below and this section will address each of these in the Production Methodology described in 2.1.

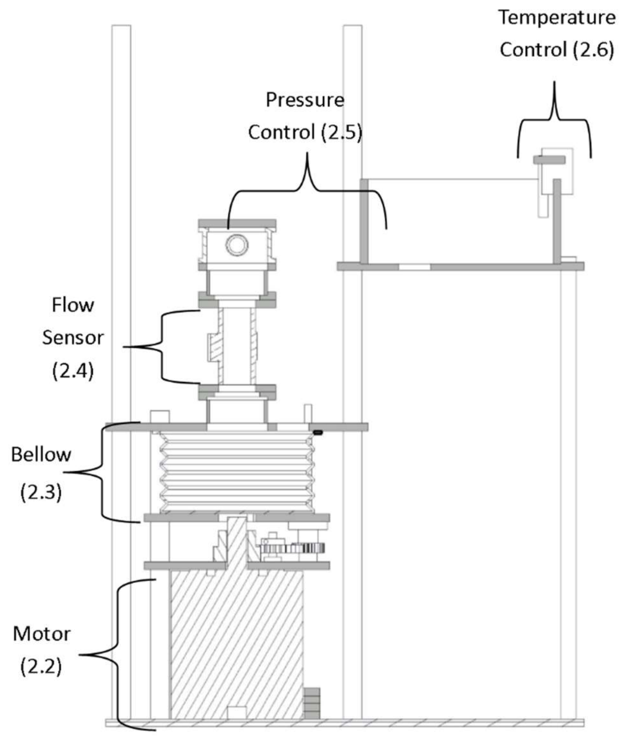

Figure 9: Diagram of Cross-section of machine and different Design and Control Systems in Report with the corresponding report sections

2.1 Production Methodology

To produce this project the machine was separated into the different components and control systems shown below;

1. Motor Control
2. Bellow Pump System
3. Flow Sensor
4. Pressure Control
5. Temperature Control

All these aspects, despite being separate things in themselves, are of course interlinked, however, this is an easy way to break the problem down into more manageable sections. Each has aspects of either electrical, mechanical or programming, and in order to build the machine separating those out a bit would be helpful. The report is also structured in this way so that it is easy to follow each decision for the project.

The methodology used to design and build this machine is a step-up methodology, as shown in the diagram below. This means that for each aspect of the project, the idea would be to go up the steps. This is helpful to visualise building on previous knowledge and allowing an agile project management technique in terms of just planning a few steps ahead every step of the way.

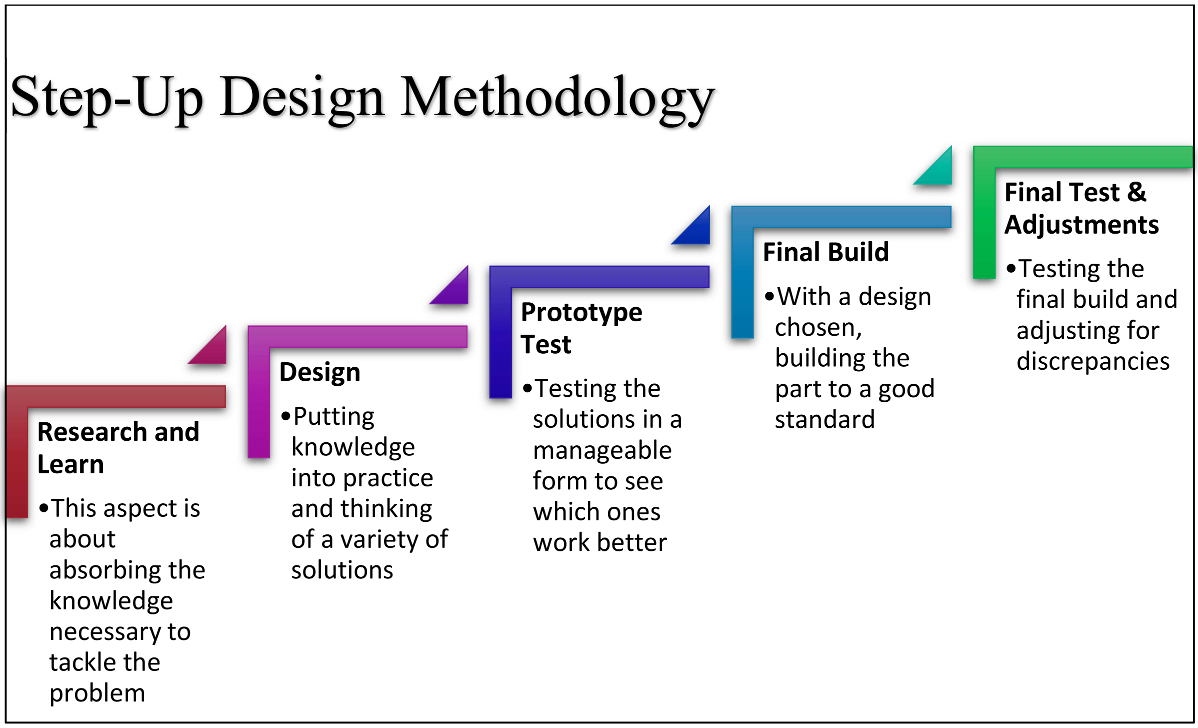

Figure 10:Diagram of Production Methodology of a Step-Up Process undertaken for the project.

2.2 Motor

The type of motor chosen for the project was a Stepper Motor(SM), as in comparison with other DC motors, a SM allows the user to control its speed and positioning through programming. This is useful for the project as the control over water flow rate is better which is vital when the flow changes in each part of the cycle. This type of motor also provides more torque to the equipment which is necessary considering how much pressure the fluid will be under when being pumped. The main advantage is also that the pump should move up and down in this instance so having a motor that can change direction is vital.

This SM will be controlled by the LabVIEW program through the myRio device. LabVIEW was chosen due to its easy to understand interface. The myRio is an embedded device which allows the control of robotic parts. To control the SM effectively, an interface of a driver is needed between the SM and myRio which is discussed next.

2.2.1 Stepper Motor Driver

To power the SM a Stepper motor driver will have to be used in conjunction with the motor and the program in LabVIEW. This is to help the input into the program and allow both coils in the SM to be controlled independently and accurately. There is a specific motor driver purchased which is compatible with myRio called the Digilent Motor Adaptor (Figure 11). As the stepper motor function will be used, the motor will need to be compatible with it, so will need to have a bipolar winding and only need 6- 16V power source.

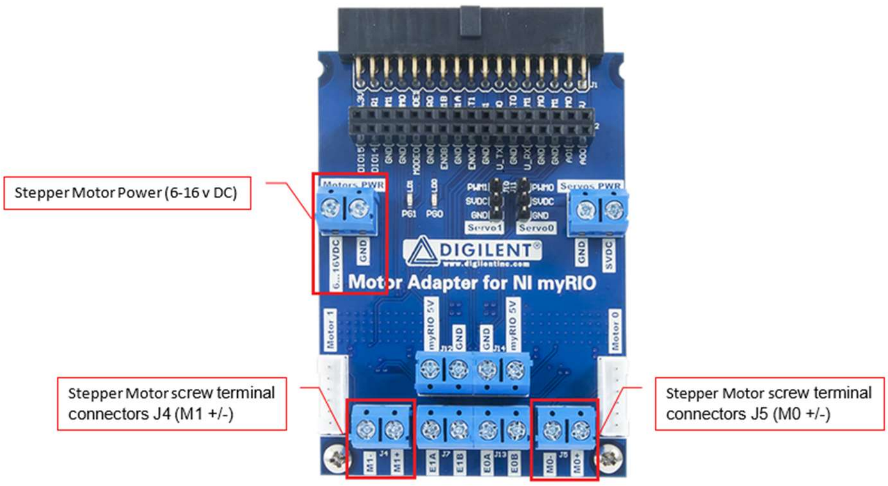

Figure 11:Picture of Digilent Motor Adaptor with labels for Stepper Motor (23)

2.2.2 Stepper Motor

In terms of choosing a motor, there were careful considerations to take forward; the amount of torque required and the compatibility with the driver. This driver was only compatible with a Bipolar Motor. This is advantageous to us in this instance as the complicated part of a bipolar motor is the driver as it needs more switches on the board, but it provides 40% more torque to the motor (24). This means that even at low speeds it will provide more torque. The other main consideration is that it must be powered between 6-16V so that the voltage can go through the driver.

The motor in the previous project was not compatible with this driver, as it required 24V and was originally wired to an Arduino. However, the torque of 1.62Nm was enough to drive the equipment so that was used as a base of the amount of torque needed for this project to work. A comparison chart was made on this basis and can be seen in Appendix 1.

Taking all this into account, the final motor chosen was the RS Pro Stepper motor (798-3640) listed in the components list. This motor provided 4.31Nm of torque, and only required 6.4V of electricity. This motor was also a hybrid motor, which means it had both wirings for unipolar and bipolar settings (25).

To find out what the wiring in the motor was, a multimeter was used to measure the resistance between the wires. The details of this process can be seen in Appendix 2 and the final diagram of the motor coils is in Figure 12. After deducing this, the wires were crimped so it would be easy to connect them to the ports while testing the program.

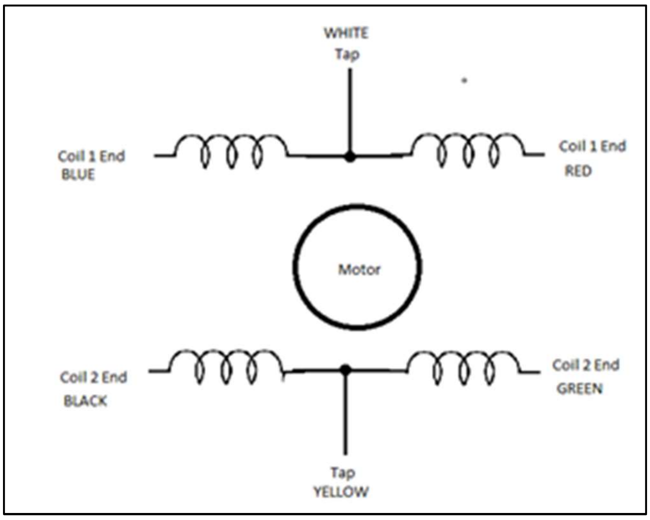

Figure 12:Diagram of coil windings in SM

2.2.3 Motor Control (LabVIEW)

The motor control would be done through LabVIEW for the easy interface. This would allow researchers to easily adjust the Pulse Width Modulation(PWM) and control the RPM of the motor. The main thing about the motor was being able to control the speed, the amount of time it was running for and the direction in which it was running.

To do this, first a WHILE loop was made to turn the motor at a certain speed, time and direction, that could all be adjusted. This was done by controlling how long it takes for the motor to go through one step (1.8°) which is a standard specification for a SM. This time could be found from the RPM which was inputted and converted as shown in Figure 14. This converter is better as RPM is more standard for researchers to use.

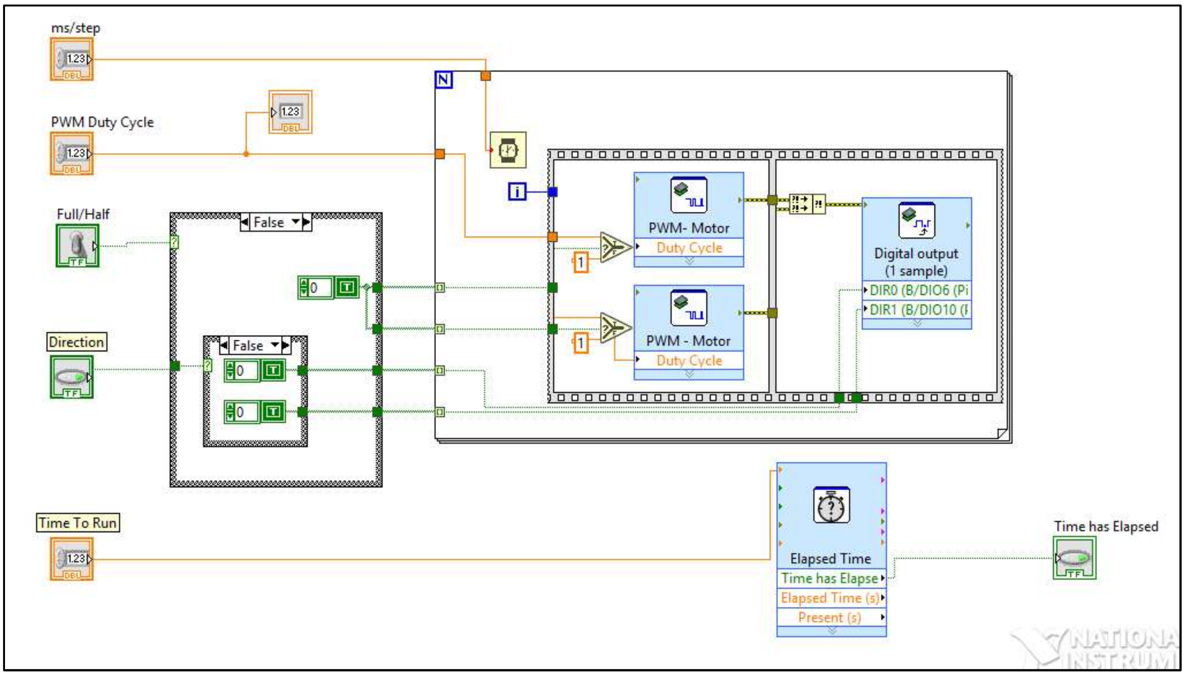

Figure 13:LabVIEW Program to turn the motor

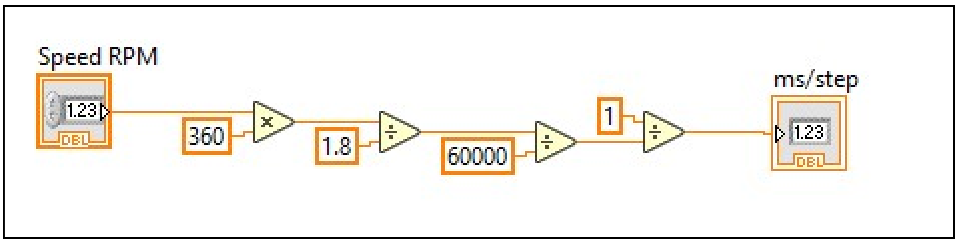

Figure 14:RPM to time per step converter

This WHILE loop was then nested in a FOR loop with arrays for each input; speed, time and direction. An array allows specific inputs to be rotated in the WHILE loop. This allowed the motor to run a certain number of speeds in specific directions and times.

This FOR loop was nested in the large WHILE loop which would allow the array to repeat just like a pulsatile cycle in the heart. This array was not included on the interface for adjustments as it is complicated, and most experiments would follow the same pulsatile flow. The LabVIEW program was also clearly labelled, so that if they did want to change it, it would not be difficult to understand which aspect they needed to change.

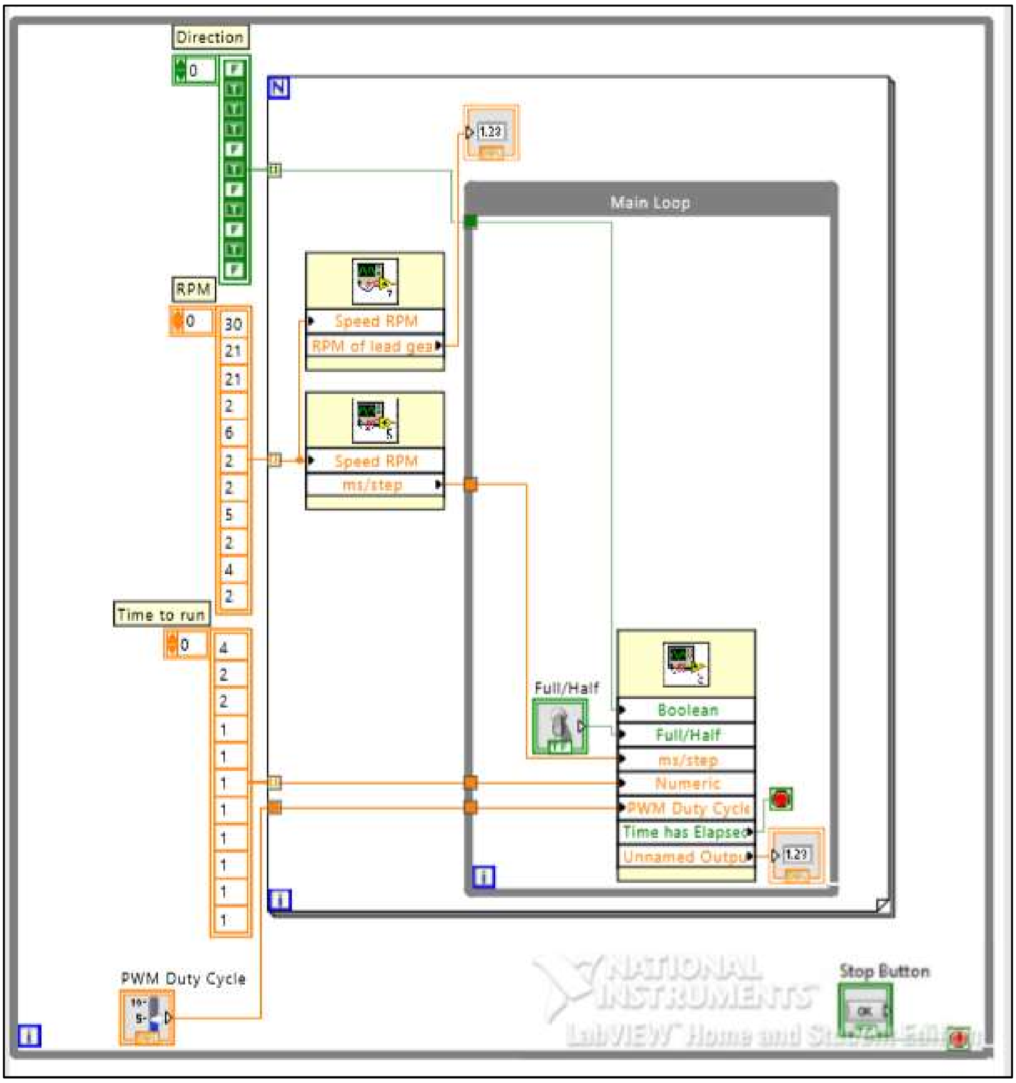

Figure 15:LabVIEW Program to turn motor in pulsatile cycle

2.2.5 Gears

The motor would power the bellow pump using precise lead screws used in the previous prototype to move the bellow platform up and down to pump the fluid. Due to the change in motors only the motor gear had to be changed. The aim was to keep the number of teeth as close to the previous one as possible so that the gear ratio was not altered between the motor gear and the lead screw gears. The previous motor gear had 20 teeth and the new one had 30 teeth. This only altered the gear ratio from 1 to 1.5 which, when considering that the bore increased from 8mm to 14mm, is an acceptable compromise. The comparison can be seen in Appendix 3. New steel lead screw gears were also purchased to substitute the plastic ones so that there would be less slip on them. The final layout of the gears in the CAD can also be seen in Figure 16. As you can see, the gears fit snugly, and the final dimensions of the new bioreactor have been changed accordingly to fit them.

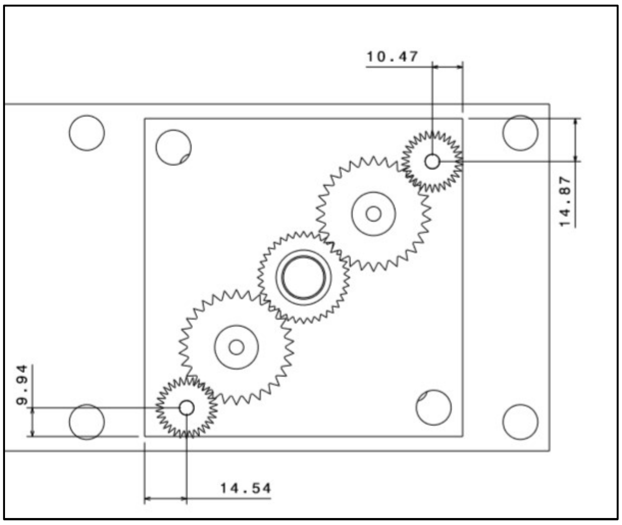

Figure 16:Top view of gears (from CATIA)

2.2.6 RPM to Flow Speed

The next step was to associate the flow speed of the liquid in the system to the RPM of the motor, through movement of the bellow. The number of teeth on the motor gear and the lead screw gear have a ratio of 1.5.

$$\frac{number\ of\ teeth\ on\ motor\ gear}{number\ of\ teeth\ on\ lead\ screw\ gear} = \frac{30}{20} = 1.5$$

The other measurement taken was the height change on the lead screw for each revolution which was measured at 24mm. Taking into account the cross-sectional area of the bellow, the volume for each revolution of the lead screw is calculated in Table 3.

$$Area\ of\ Bellow = \pi \times 50^2 = 7854\ mm^2$$

Table 3: Table showing the volume pushed for each revolution of the lead gear and motor gear respectively

No of Revolutions of Motor Gear	No of Revolutions of Lead Gear	Height Change of Bellow (mm)	Volume Moved (mm^3)	Volume Moved (ml)
0.667	1	24	188496	188
1.333	2	48	376991	377
2.000	3	72	565487	565
2.667	4	96	753982	754
3.333	5	120	942478	942

As the lead screw is only 80mm tall, the program should not make it spin for more than two revolutions of the motor gear. To change the flow-speed, the rpm should change and control how quickly it should complete two revolutions.

$$Time\ for\ 2\ Revolutions = \frac{60\ \times 2}{RPM}$$

$$Flow\ Rate\ \left(\frac{ml}{s}\right) = \frac{565\ ml}{Time\ for\ 2\ Revolutions\ (s)}$$

This equation was used to plot both variables. This can be seen in the line graph below. The equation using all the points is;

$$Flow\ Rate = 4.7123 \times Motor\ RPM + 0.006 \qquad (Eq. 1)$$

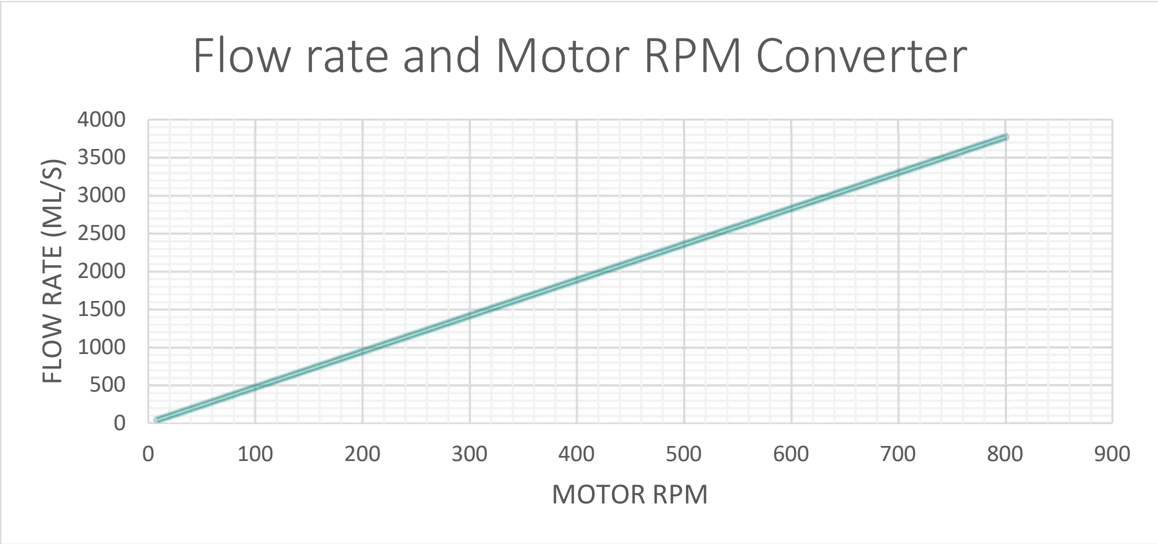

Figure 17:Graph showing the relationship between RPM and Flow Rate

In the end the idea is to be able to create a pulsatile flow that mimics that in the heart and cycles around. To plot this, I used a general plot based on that measured in vivo on the AV and applied Eq.1 to calculate the corresponding motor RPM. This was then inputted into the array of the motor LabVIEW to make the motor spin at different speeds in different directions and really simulate the heart's flow speed.

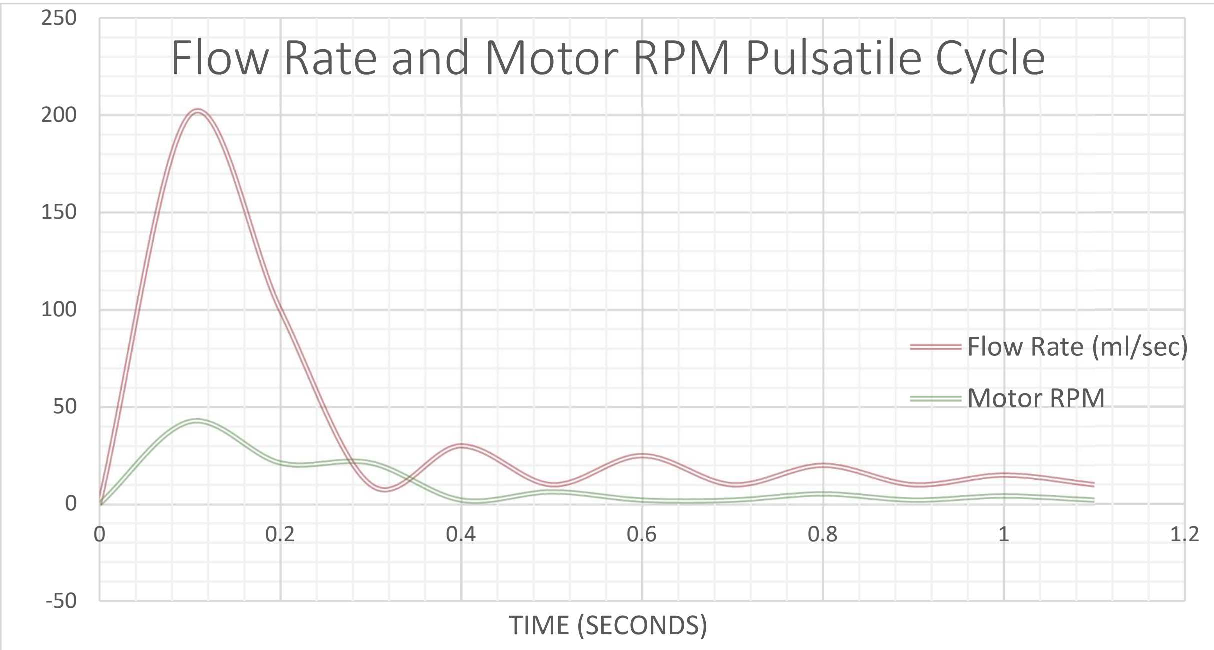

Figure 18:Graph showing the pulsatile flow rate programmed into Motor RPM to create a pulsatile flow from the motor control.

2.3 Bellow for Pump

The bellow pump was chosen as per the comparison in the previous project (21) as it was the best option to fulfil the requirements below;

- Up to 210ml/s volume change
- Durable
- Biocompatible
- Compact

The previous bellow was made from latex, which is not biocompatible, and was glued to the acrylic, which made it hard to clean. The dimensions were still adequate and so the same mould could be used. The calculations for this stroke and volume rate flow speed have already been discussed and are enough to mimic the biomechanical heart conditions. The choice of the bellow pump is also ideal for this project as it is similar to the hearts' structure and how it pumps blood.

2.3.1 Bellow Manufacturing

To make the bellow biocompatible, as it would be in regular contact with biological fluids, silicone was chosen. This is because the material can be strong and flexible, but also does not support bio-microbial growth, which is essential when designing medical equipment that needs to be sterile. Outsourcing the manufacturing of this would have resulted in costs of above £500 so it was manufactured in-house.

The silicone used was a standard T25 Silicone Rubber mixed with its catalyst (26). This would give the flexibility and durability needed. A thixotropic additive was used to allow the mould to be skin moulded and to avoid a runny mixture and make a smoother finish (26).

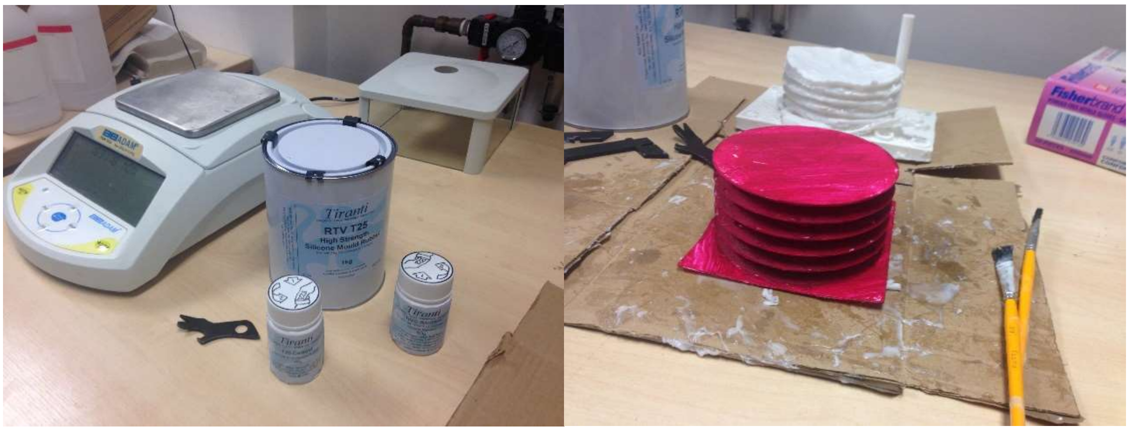

Figure 19:Picture of Bellow Moulding setup

Figure 20:Photo of final bellow

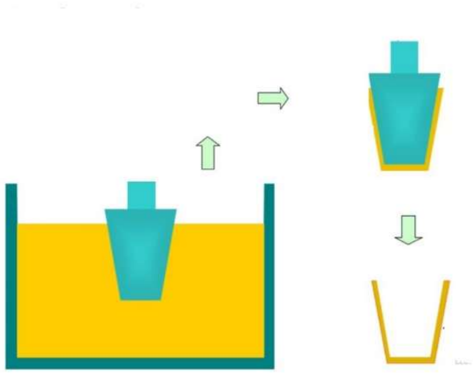

Figure 21: Diagram of Dip Moulding Process (27)

A dip moulding process was used to make this. The silicone and catalyst were mixed together thoroughly. Then it was spread on the mould to create a 3mm thick bellow, which was then pushed out of the mould.

To attach the mould to the machine, the border was sandwiched between two acrylic sheets that were screwed together. This would mean that to clean it, taking it apart would only require a few minutes and it could be set back up.

Figure 22:Picture showing Bellow attached in machine

2.4 Flow Sensor

2.4.1 Flow Sensor Choice

The main biomechanical condition we are mimicking is the flow rate. This would be produced by the bellow pump and would also be monitored live through the LabVIEW program. The flow speeds this sensor needs to read are between 0-250ml/s. This equates to up to 15litres/min. The following flow speed sensor was kept from the previous project (21). This was a flowmeter based on the hall effect, which uses the change in magnetic field to recognize the flow passing through. Technical details on this effect can be seen in the Appendix 4. According to the product details the accuracy for a continuous flow was 10% and it could measure flow rates from 2-45liters/min (28). This means it can sense speeds as low as 33ml/s, which is small enough for the use of this bioreactor.

2.4.2 Calibrating and Wiring

The wiring of the flow sensor can be seen in Figure 23. This was done on a breadboard and a 10kΩ resistor was used.

The next step in the prototyping stage was to calibrate the flow sensor and program to record the flow rate on LabVIEW. In terms of the calibration process, the flow meter was set up with a tap (continuous fluid source) and a measuring cylinder underneath. The program was run on LabVIEW to see what the fluid flow was, and matched against the readings taken manually. This ranged between turning the tap on at 4L/min and 1L/min. Due to the fact that the tap was not completely uniform in the way it would run, the flow speed did oscillate in the readings, but when an average was taken, it matched the manual readings by the end of the process. Figure 24 shows how the equipment was set up to take these readings.

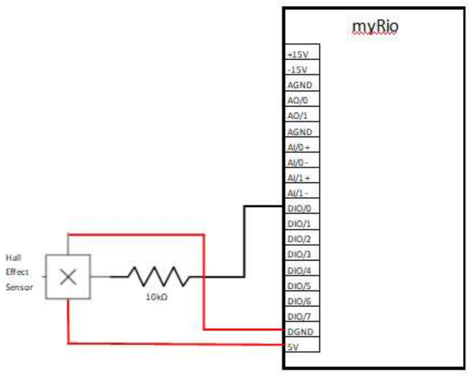

Figure 23: Wiring of flow sensor to myRio

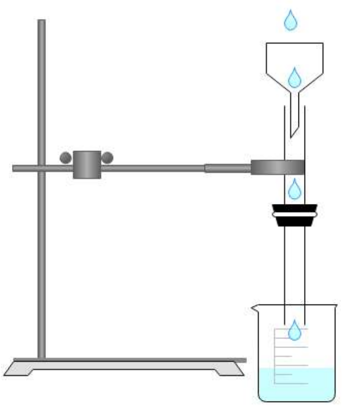

Figure 24: Diagram showing the experimental setup when calibrating the flow meter.

2.4.3 Programming

To program the flow sensor, the signal coming in first had to be identified. This signal was digital. By referring to the Appendix 4 on the hall effect, it was clear that a change from true to false was caused by the rotor spinning once. Referring back to the product details: *"The cumulative flow pulse conversion ratio 1L Water=477 pluse +/- 10%"* (28).

Due to this assumption, we could deduct the volume of fluid going through for around 1 pulse as 2.10ml.

By referring to the program documentation, if the signal had changed from either True or False to theopposite, then the time elapsed was recorded and reset. This was then averaged with 7 other readings to give a good cross-section of what the flow rate was without as much variation. Then 2.09ml/pulse was divided by the time and converted to give a value in L/min. This was then displayed live on a chart while the program was running, and then a final graph of the whole run time was displayed. Many more charts were included in the calibration program shown in Figure 25 to check for issues and the final program is in Appendix 6.

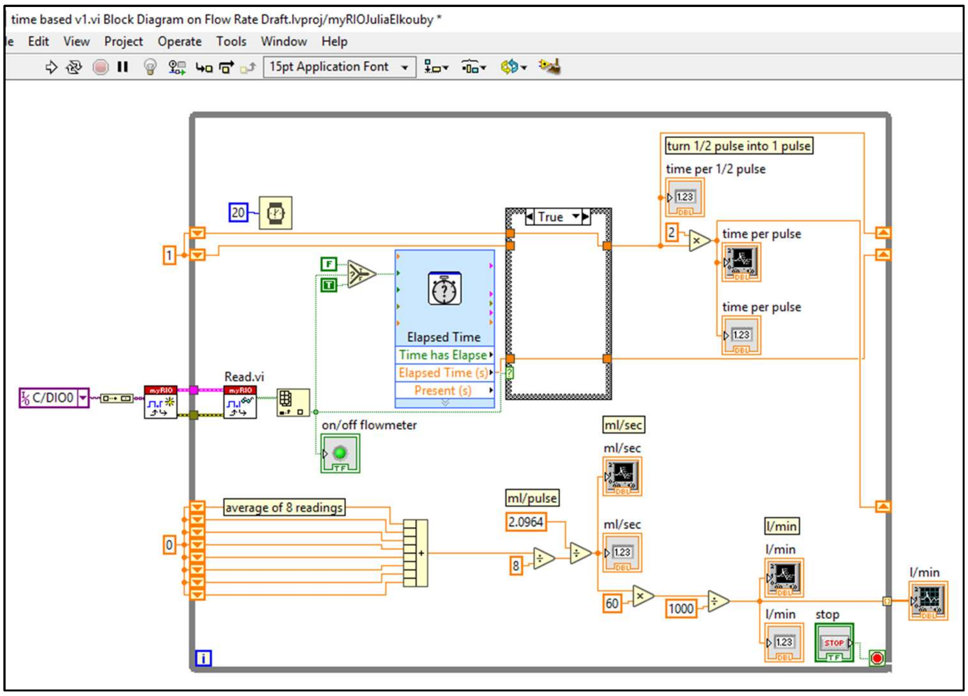

Figure 25 LabVIEW program to calibrate and test the flowmeter.

2.5 Pressure

2.5.1 Pressure Calculations

One of the primary biomechanical properties of an AV is the pressure it is under. This was one of the main aspects that should be simulated to create realistic heart conditions, and is also much of a step up from the previous project's pressure of a pulmonary heart valve. The pressure difference can be defined mathematically, as shown below, with resistance being the variable that is needed to increase in proportion to the pressure difference needed;

$$\frac{Pressure\ Difference(\Delta P)}{Flow\ Rate(Q)} = Resistance(R)$$

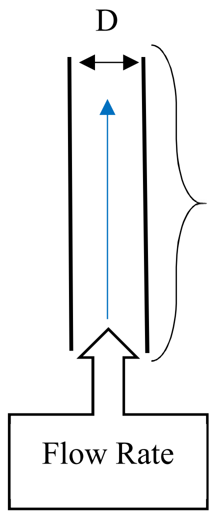

Figure 26: Diagram showing how the fluid will flow up a tube of a certain length to give a pressure difference equal to that of an aortic heart valve.

The resistance can be increased by changing the tubing coming off the heart valve and its properties. These are defined by Poiseuille's Law, which shows how the resistance(R) of a tube can be controlled by changing its diameter(d) and length(L);

$$R = \frac{128\rho L}{\pi d^4}$$

In the previous project, the tube coming off the valve made a pressure difference of 2.5mmHg, which is just 10% of the pressure needed for a pulmonary heart valve. This was done for ease of the first prototype (21). However, as an AV supports up to 120mmHg of pressure, the tubing should be able to be swapped and reach the best part of this pressure. To do this, different spreadsheets were made to calculate the resistances for all the different lengths of tubing with a few components kept constant to calculate the resistance accurately. If only one tube like in the previous project was used with a 10mm diameter, the length of the tube to reach 120mmHg would have to be 26.1m long, which is unreasonable. A solution would be to restrict the diameter of the tube so that it is smaller. However, this encounters another issue which is that making the tube too small could create turbulence in the flow as the tube is much smaller on diameter than the heart valve.

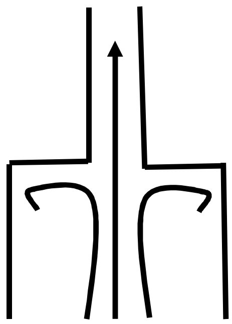

Figure 27:Diagram showing how a radical change in diameter can cause the flow to become turbulent.

To avoid this issue the tubing would be made from multiple tubes going from a larger diameter to a smaller one. This is of course less favourable in terms of making the necessary pressure difference in a short length of tubing, but is still better in terms of having a much more laminar flow through the valve.

Table 4:Table showing the length the tubing would have to be for the necessary pressure difference or the necessary diameter of tubing for the length to be around 20cm

Pressure(mmHg)	**Resistance needed(Pas/m^3)**	**Diameter(m)**	**Length(m)**
15	11853689	0.01	3.269
		0.005	0.204
120	94932909	0.01	26.180
		0.003	0.212

As the tubing is attached one after the other the resistances are summed like in a series circuit. To calculate these lengths the density of liquid used was that of water as that is the liquid that would be used for initial testing. Since the pressure is not created by the height, the tubing could be made by flexible tubing which would be easier to move around. However, the limit of length should be around 35cm so that it is not too cumbersome and expensive. To make the tubing easily changed for different pressures, the initial tubes of larger diameters (0.010m, 0.008m, 0.005m) were kept of constant length and only the tubing of diameter 0.003m is changed for each pressure (Figure 29). The final lengths for each pressure can be seen below.

Table 5: Table with summary of different tubing lengths for the different pressures.

	Resistance(Pa.s/m^3)	***Length of tubes(m)***
120mmHg	*94947681*	*0.201*
15mmHg	*11868460*	*0.165*

This would allow researchers to easily customise the pressure they want their experiments to run at by just adjusting the length of the 3mm tubing. This would eventually mean that they could test different types of heart valves that are under different pressures without changing any other moving part. This would be calculated and adjusted beforehand using an Excel sheet that has clear inputs and is provided in Appendix 5 which is also on the memory stick attached.

Table 6:Table showing the lengths of 3mm diameter tubing needed for each pressure difference.

Pressure (mmHg)	**Length of 3mm Diameter tubing (m)**
120	0.202
110	0.184
100	0.166
90	0.149
80	0.131
70	0.113
60	0.096
50	0.078
40	0.06
30	0.043
25	0.034
20	0.025
15	0.016

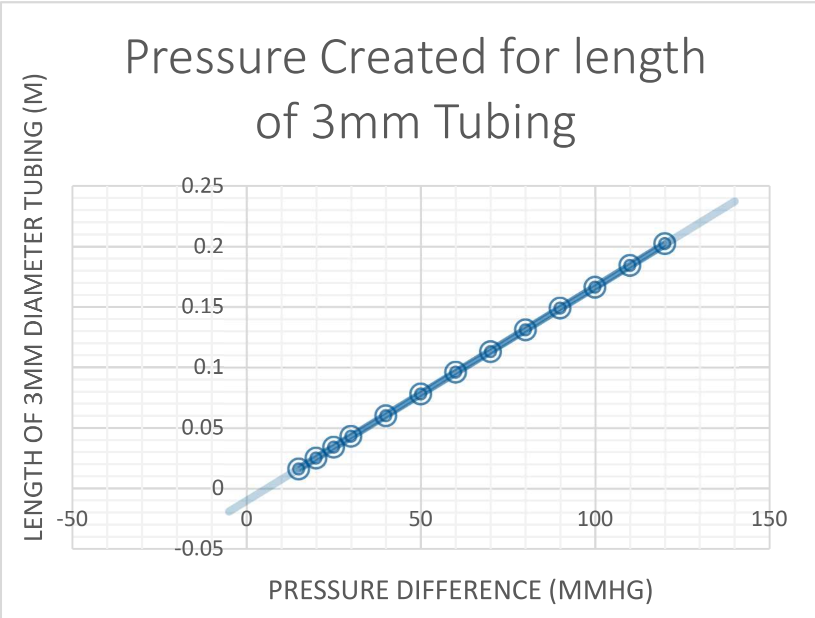

Figure 28:Graph showing the lengths of 3mm diameter tubing for each pressure.

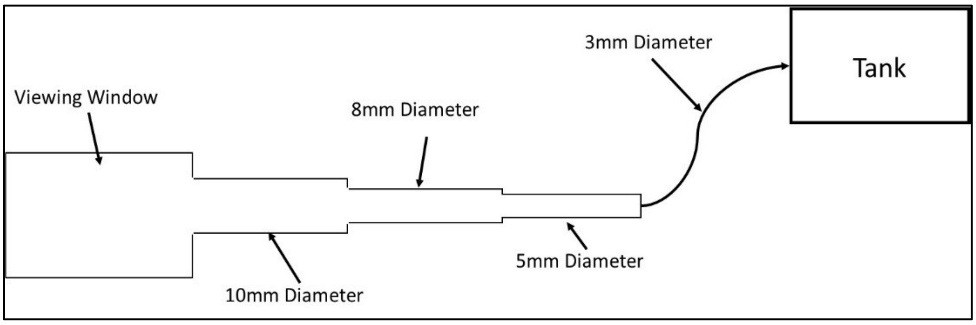

Figure 29:Diagram of Tubing

2.5.2 Tubing Manufacturing

For the tubing, a flexible PVC tubing was chosen, which could be custom cut from seller. The next requirement was attaching them so that the tubing stayed watertight. To do this, attachments were laser cut from acrylic in the same way as the rest of the machines, so that the tubing could be screwed together easily(Figure 30). This allows it to be easily customisable by any researchers at UCL, as there are readily available laser cutters for extra attachments.

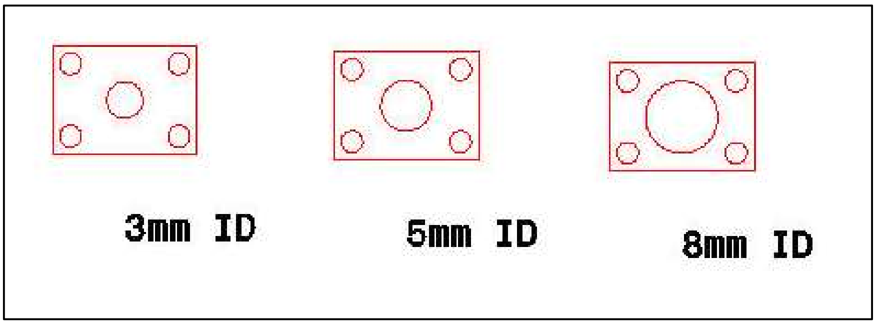

Figure 30:Laser cut of Tube Attachments

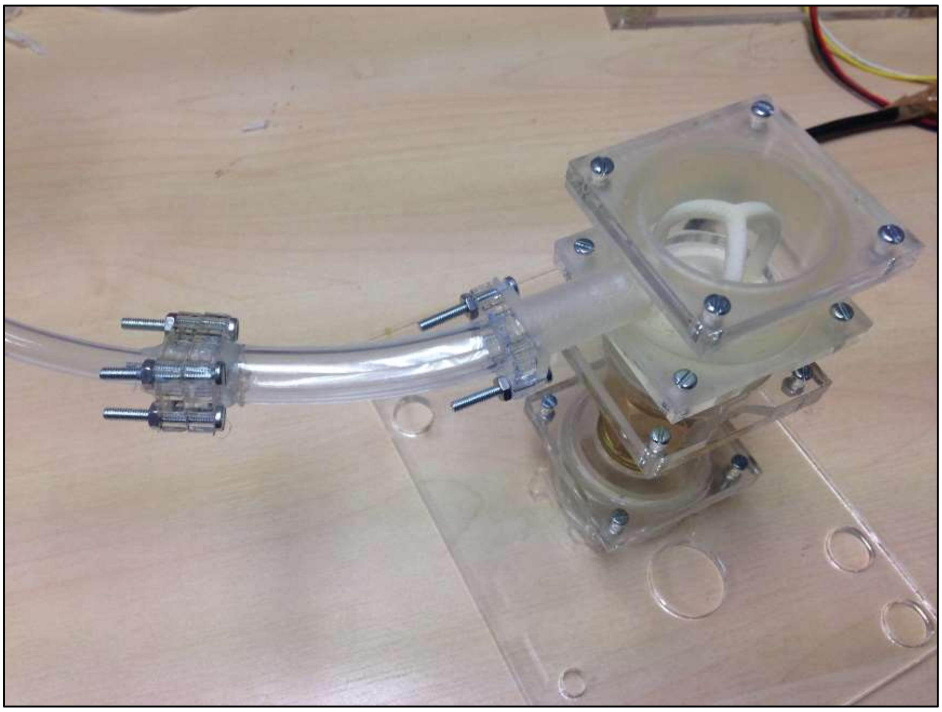

Figure 31:Picture of PVC Tubing attached with attachments

2.5.3 Viewing Window Design & Manufacturing

One of the main aspects of the new design with the tube coming off from the side meant that a clear viewing window could now be implemented as required. The design of the main body was 3D printed out of standard resin in the SLA Printer. This meant that there was no risk of the tube coming off from the side being manufactured in a way that would not be watertight as cutting acrylic, which is brittle, can cause cracks. Although this material is slightly opaque, the top was made from clear Perspex acrylic and put together with screws and bolts. This gives the researchers a clear top view of the valve and means that it is much sturdier and watertight as it is made from the least number of parts as possible.

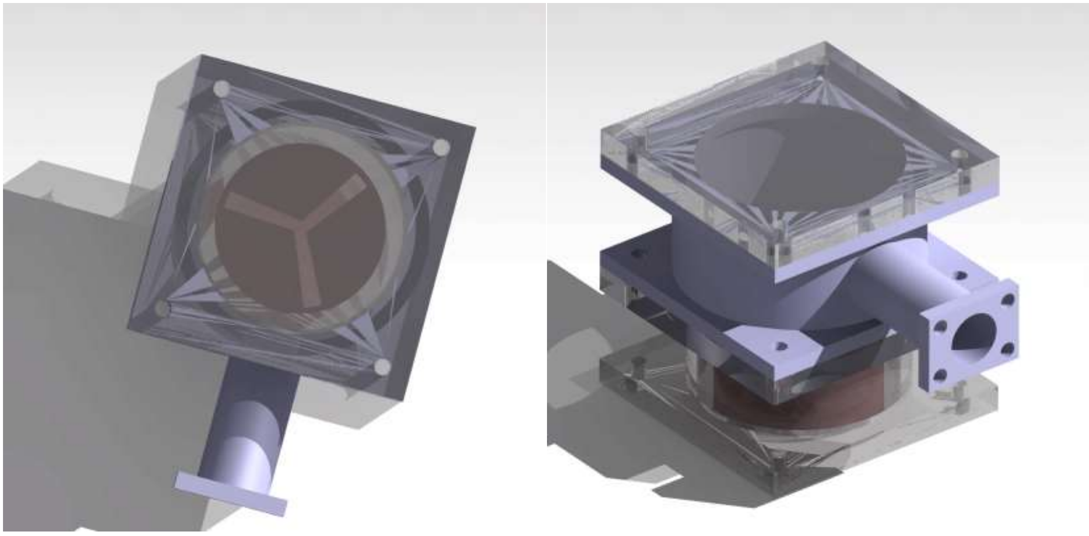

Figure 32:Rendering of Viewing window component.

2.6 Temperature Control

2.6.1 Temperature Sensors

The main choice in temperature sensors was choosing between a thermistor and a thermocouple. The difference between these is mainly the fact that the temperature ranges of thermocouples are much larger than thermistors (29). Thermocouples are also generally more durable as they can withstand more pressures and temperatures, and are used in a wide variety of engineering applications. The main criteria for this use was the durability, as it would be in constant contact with water; therefore, thermocouple was chosen.

The issue with thermocouples was the fact that the signal was in mV which would be hard to pick up with the LabVIEW. The solution to this was to use an Operational Amplifier (Op-Amp) to amplify the signal. This, used with resistors of 100kΩ and 1kΩ, would create the gain of 100 which is significant enough to see a change of signal in LabVIEW.

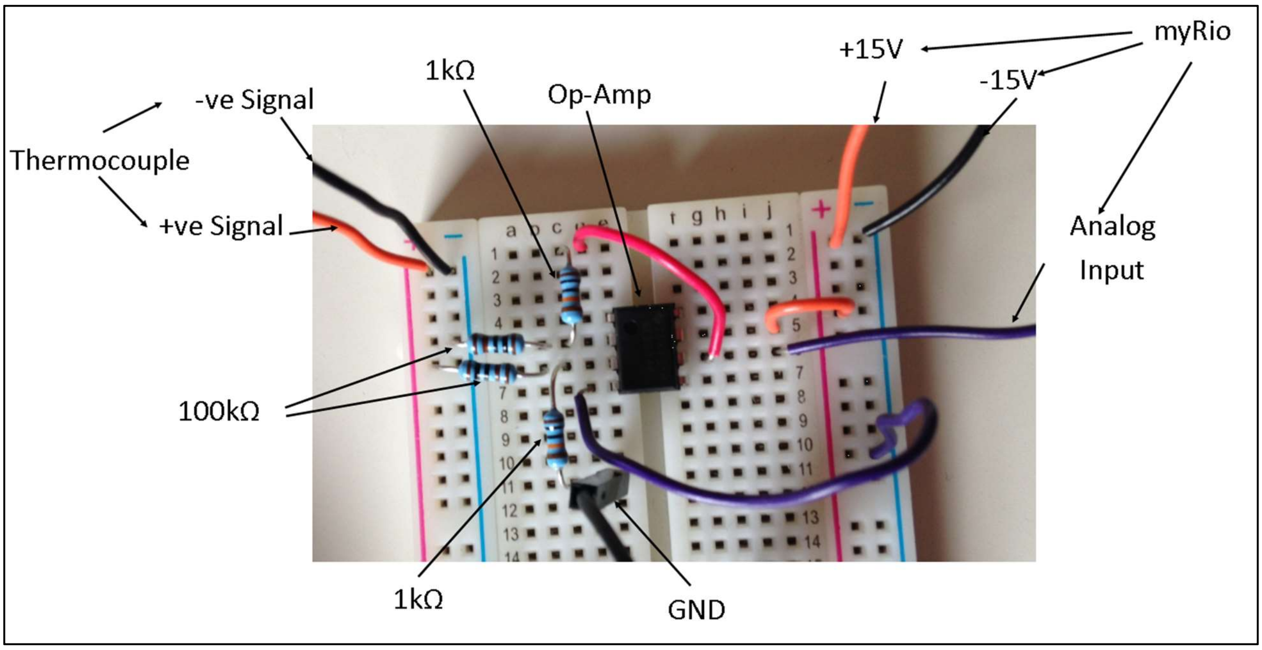

Figure 33:Wiring of the Op-Amp in conjunction with the thermocouple

2.6.2 Water heater

Many different types of heaters could be used for this project to heat up the fluid. To rank these and choose the most suitable one, a pairwise comparison was made. The criteria included the heater being waterproof and being able to fit in, as well as be powered easily by mains electricity. Biocompatibility would also be an issue in the future, but for the moment it is less important than getting the other parameters.

Table 7: Table of ranking of criteria for water heaters and pairwise comparison of the heaters

	Waterproof	Fit in Dimensions	Small Voltage	Price	Biocompatible	TOTAL	RANK (5-most important and 1 least important)
Waterproof		0	1	1	1	3	4
Fit in Dimensions	1		1	1	1	4	5
Small Voltage	0	0		0	0	0	1
Price	0	0	1		1	2	3
Biocompatible	0	0	1	0		1	2
Ceramic Heater	2	4	3	2	5	47	2
Band Heater	1	2	2	4	4	36	4
Cartridge heater	3	5	5	5	3	63	1
Water heaters	4	1	1	3	2	35	5
Liquid Heat Exchangers	5	3	4	1	1	44	3

As you can see from Table 7, the Cartridge Heaters were chosen for their suitability in the project. This can be clearly seen in the pairwise, and after reading more on the subject, they are also widely used in research, so are clearly the best choice.

2.6.3 Temperature Control Programming

In order to calibrate the thermocouple, an average value was produced by the LabVIEW program once it was wired up to understand the signal. It was calibrated by holding the thermocouple in water with a thermometer to know what temperature the thermocouple was measuring and so plot the signal and temperature on a graph(Figure 35). Using the regression line, I was able to make a LabVIEW program that converted the signal of the thermocouple into the temperature it equated. As can be seen from the regression factor of 98%, the variability results in this final equation;

$$T = \frac{V + 3.0251}{0.0142} \qquad (Eq. 2)$$

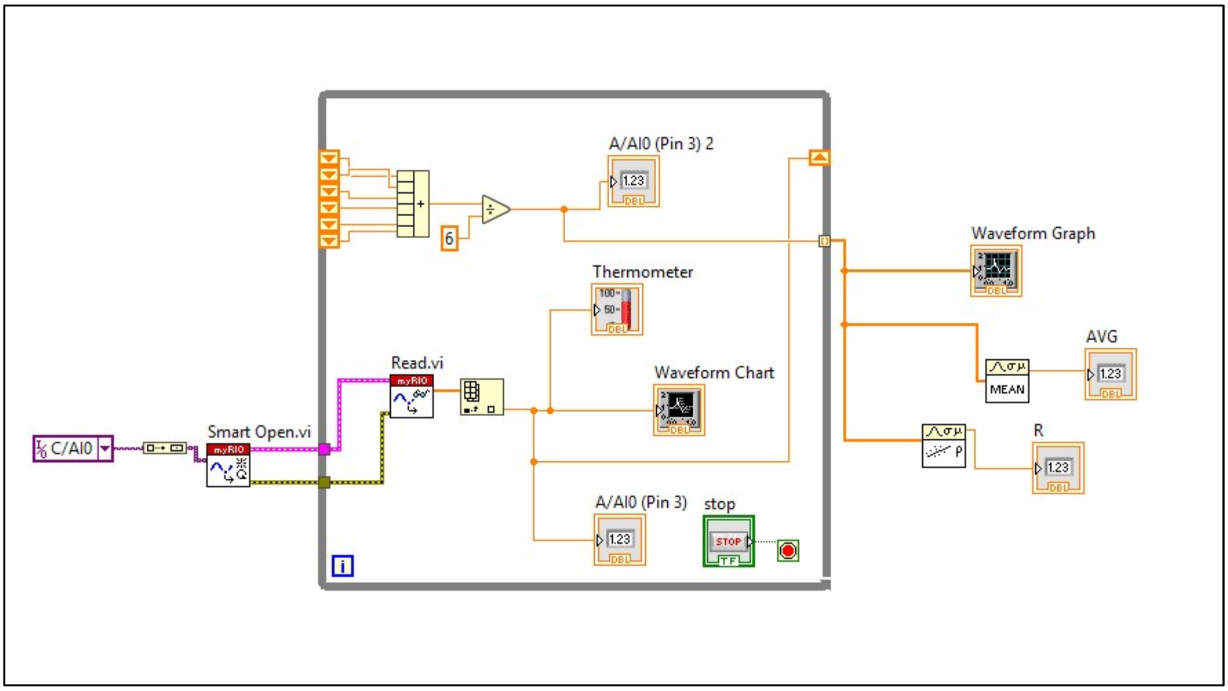

Figure 34:LabVIEW program used to calibrate the Thermocouple

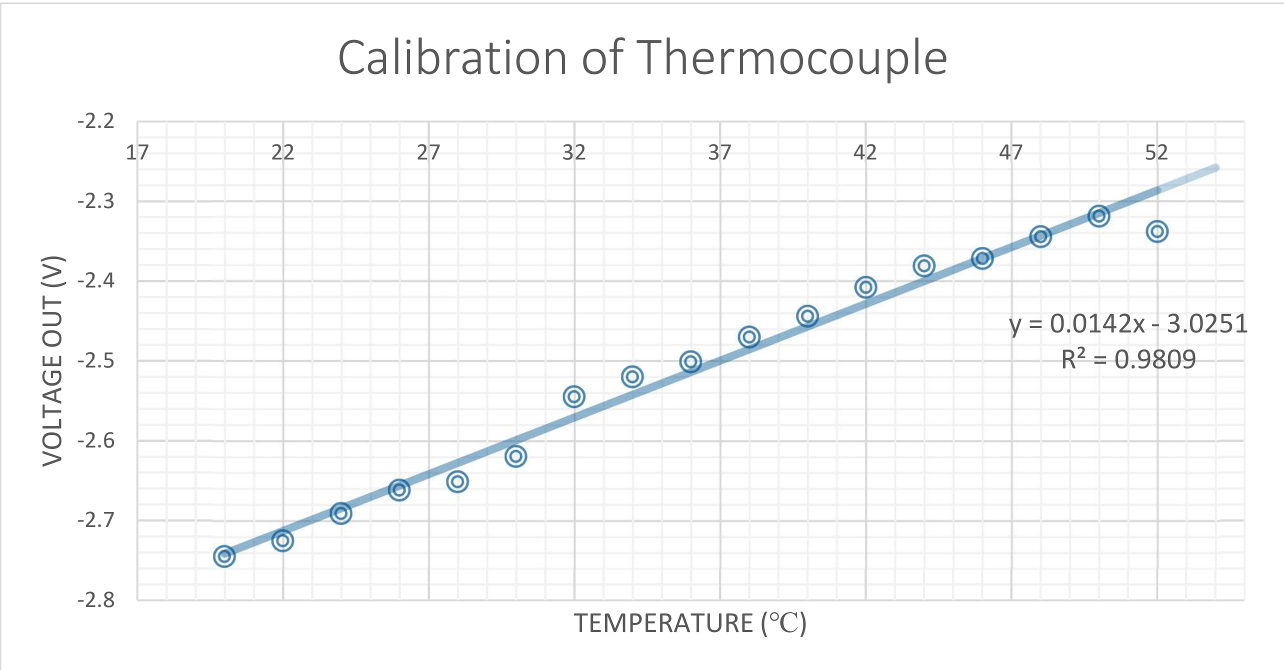

Figure 35:Graph showing the relationship between voltage signal and the temperature of the thermocouple

Using Eq.2, the LabVIEW program in Figure 36 was designed to control the temperature of the fluid. This was done using a MOSFET, which would turn the signal on/off depending if the temperature read was below/above the required temperature. This would take a while to stabilise, but a simple PID control could be implemented in a later version of the project.

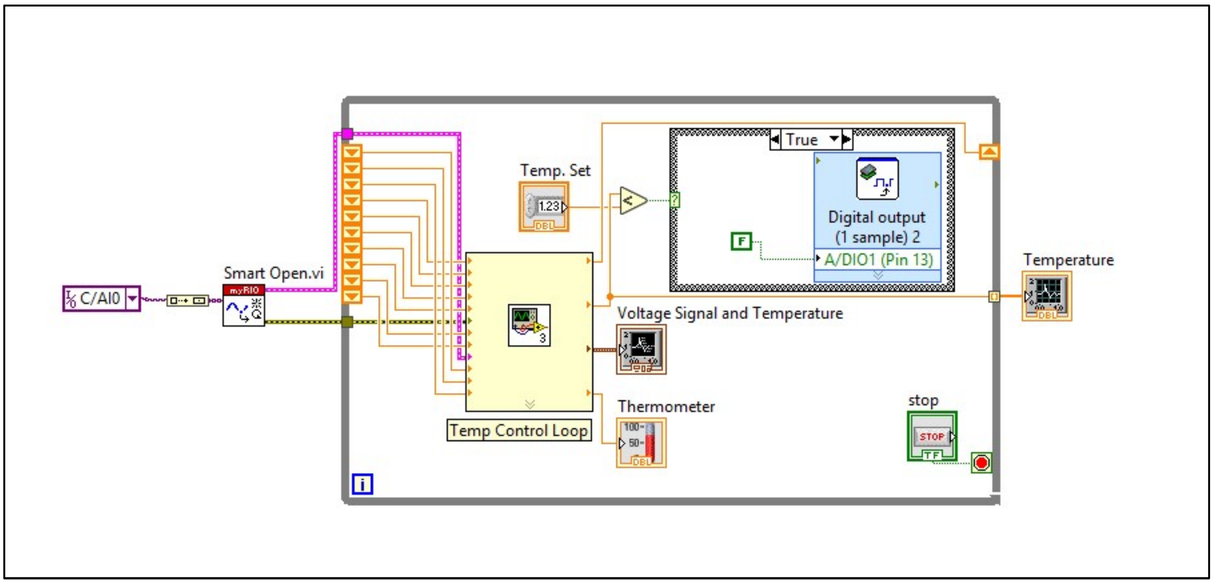

Figure 36:LabVIEW program to control the cartridge heaters

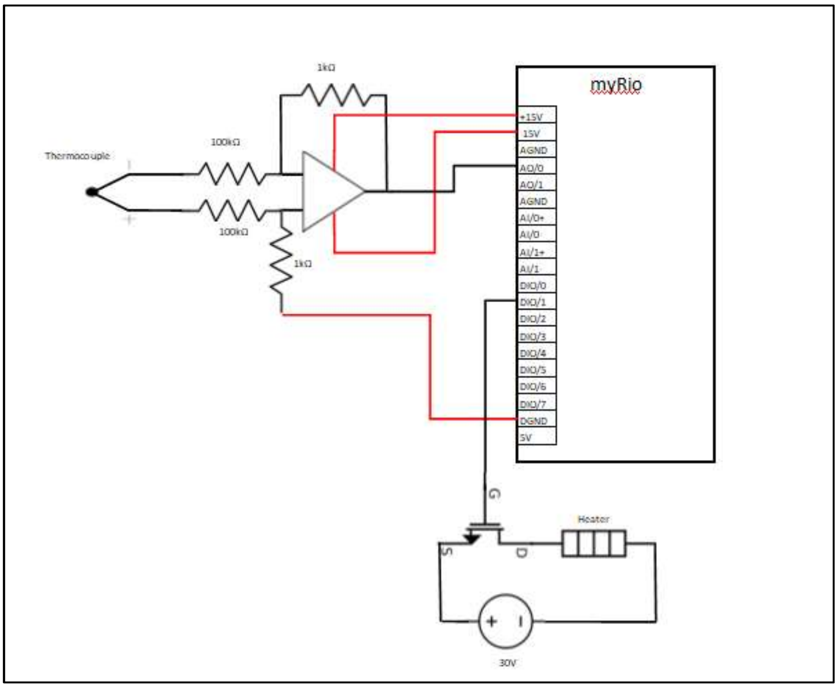

Figure 37:Schematic of Temperature Control Wiring

3.0 Final Design

3.1 Hardware

One of the main requirements of the hardware was that it would be modular so it could be easily taken apart and put back together. This was to make sure that the machine could be easily cleaned, which is very important when experimenting with biological fluids. This is also important in research, so that each researcher can easily customise each experiment and add sensors depending on what they are testing.

To do this, the idea was to create a frame with some pillars. This would then allow each block to be suspended on the pillars. The whole machine would also be put together with screws and bolts which are easy to take apart and put back together. The drawings of the hardware can be seen in Appendix 7. All the acrylic platforms were laser cut to size and attached accordingly.

Another aspect was making the machine water tight. To test this, each component was tested before on their own and O-rings were used between components to minimise the risk of leaks during a malfunction.

The other aspect considered was the fact that the motor vibrating would wear down the hardware and make it vibrate, so it would be less efficient. To help with this, a rubber pad was put beneath the motor as well as cushion pads under the pillars of the machine to make it as sturdy as possible.

3.2 Electrical Hardware

In terms of electrical hardware, everything was to be controlled by myRio and the LabVIEW on a laptop. The wiring can be seen in Figure 38 so that as much of the myRio is utilised. The Op-Amp and flowmeter can both be powered by the myRio directly and the motor and heater are powered by a power source from the mains.

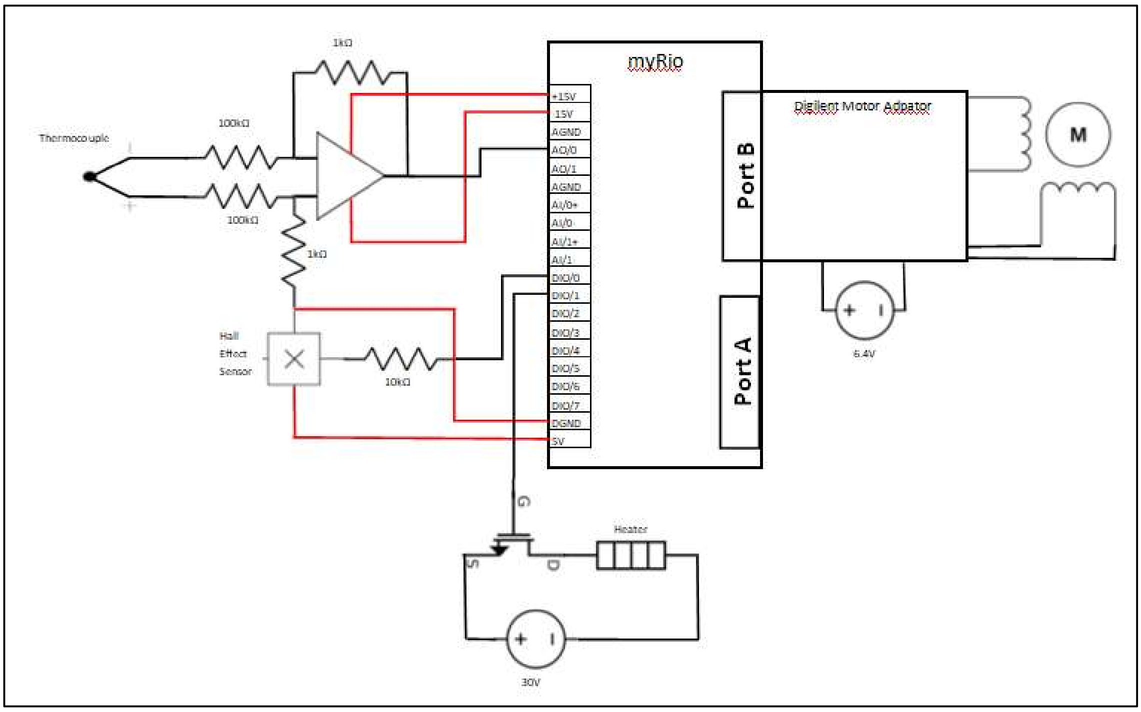

Figure 38:Diagram of MyRIO wiring and whole schematic diagram

3.3 Programming

The final program was all done in LabVIEW, which has an easy interface and can be used easily by researchers of all backgrounds. The documentation and all the diagrams of each subVI[1] can be seen in Appendix 6. The main interface is below and, as can be seen, the whole interface fits easily onto one screen and has clear instructions for the running of the machine. This is easy to understand and easy to adjust depending on the experiment.

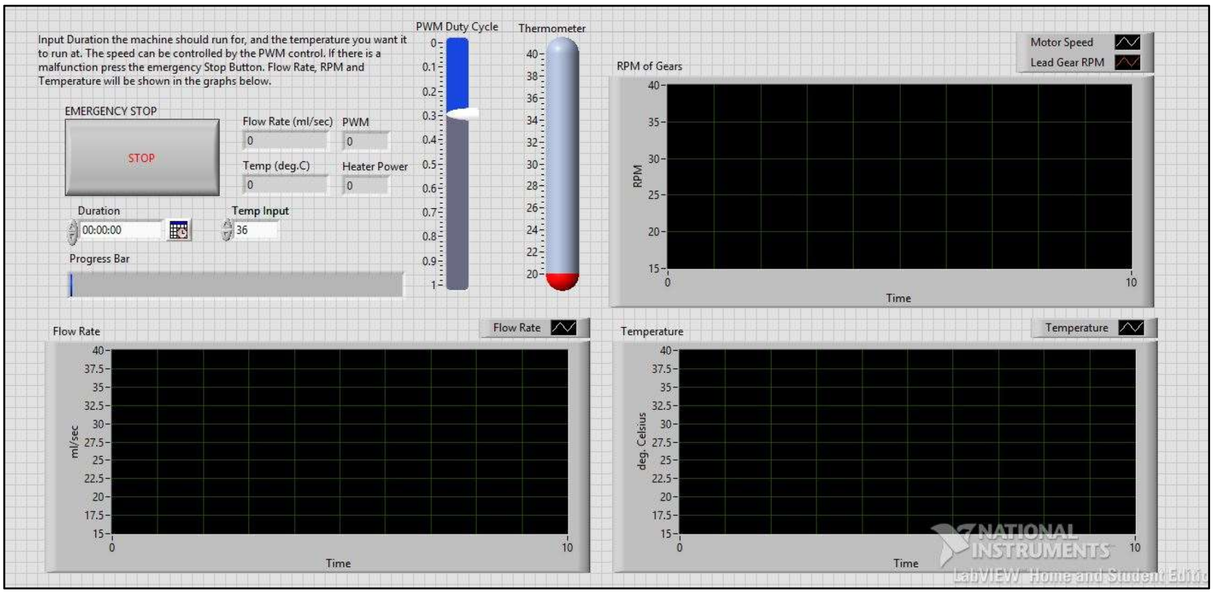

Figure 39:LabVIEW Interface

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[1] subVI refers to a LabVIEW program that is summarised to a single icon. Similar to the use of simple methods in a large program.

The block diagram behind it is all labelled and parts that can be adjusted easily are not in any subVI so that it is easy to change. This program can be run on myRio and a copy of this is attached in the memory stick provided.

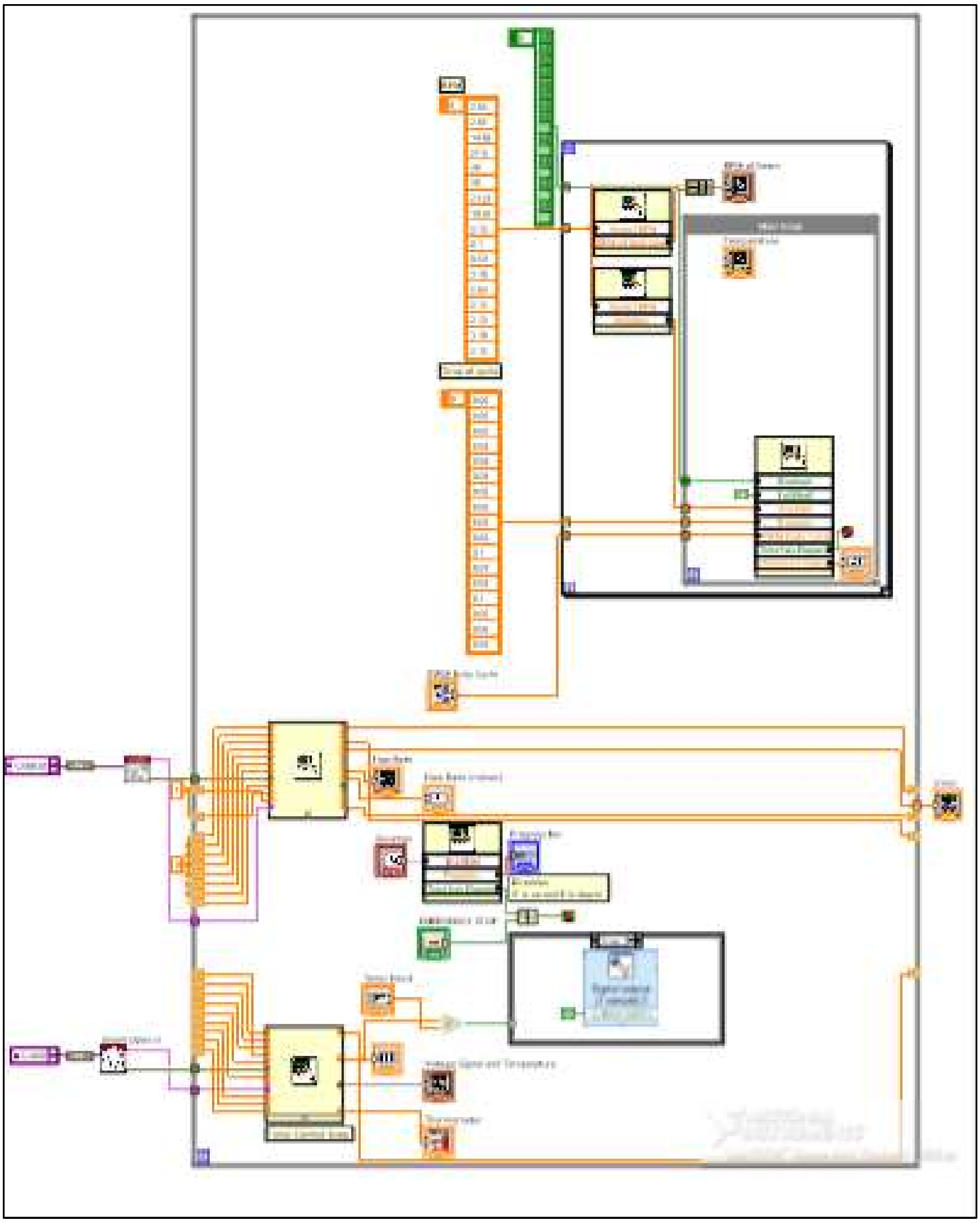


Figure 40:Block Diagram of Main Program

3.4 Building & Testing

While building the project as described in the previous sections, it was tested throughout against the requirements. This last section will show an analysis of the performance against the original requirements.

1. Flow Rate

The flow rate was calibrated and tested with water as you can see in Figure 41. A few different flow rates with different response times were measured. The average response time is 0.483 seconds which means that the readings are only ever less than half a second delayed. This creates a very stable flow rate signal, which is good as the pulsatile flow will change constantly.

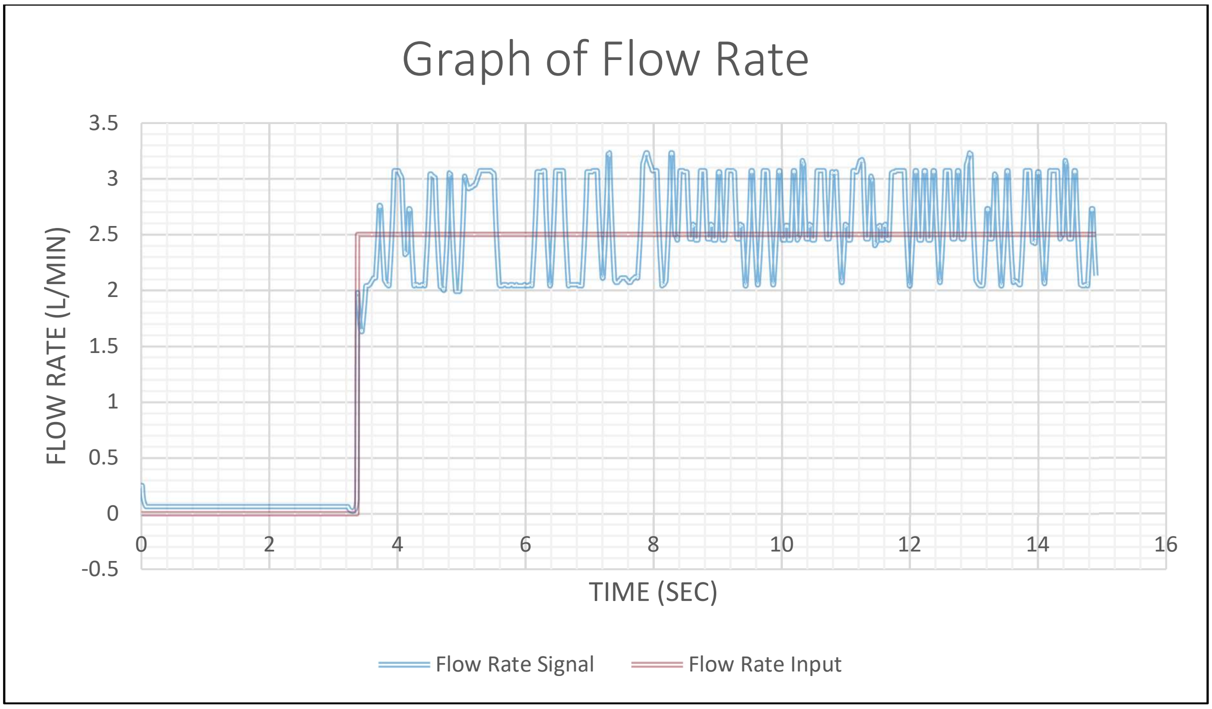


Figure 41:Graph showing flow rate when the whole machine is running.

2. Temperature

The temperature equipment was held up in the tank as seen in Figure 42. The thermocouple was calibrated as discussed before. To test the control system with the cartridge heater the temperature was measured while one and two cartridge heaters were on. These were powered with an external power supply at 30V each from mains electricity. As can be seen from the graph the time it took for the water to heat up the by 1°C was halved when both cartridge heaters were used.

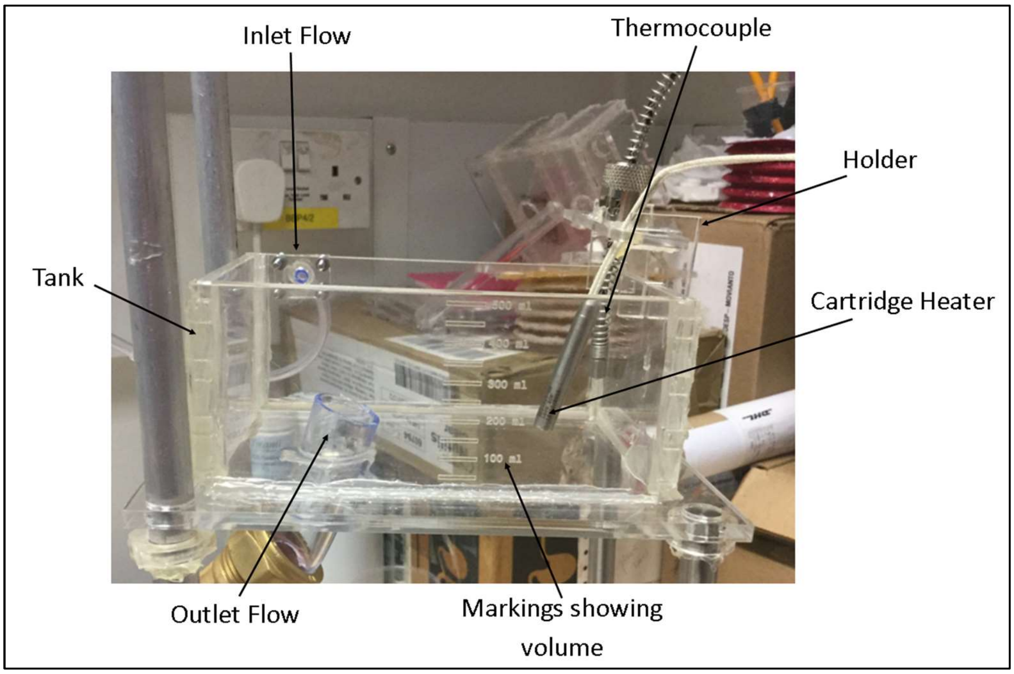


Figure 42:Picture of tank with labelled components

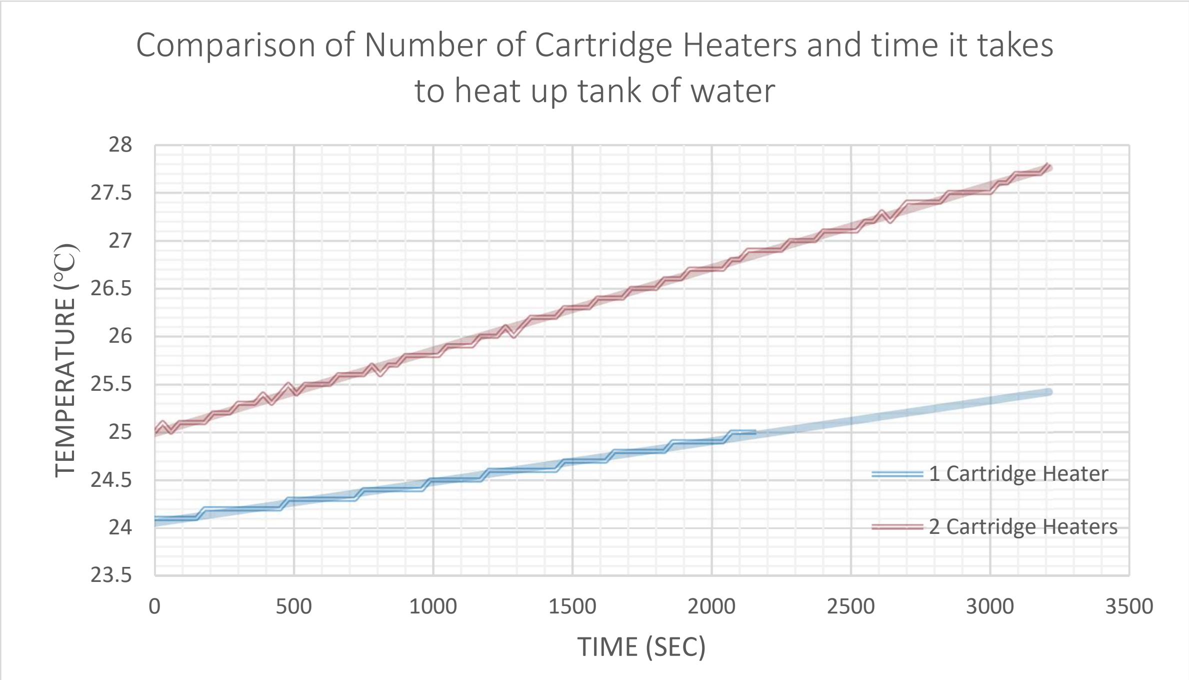


Figure 43:Graph showing the response time to reach the right temperature it was set at

This shows that the cartridge heaters are not as powerful as expected. To improve upon this more cartridge heaters could be used to heat the water up more quickly. Since the time it takes for the fluid to heat up is inversely proportional to the number of cartridge heaters, this can be plotted as in Figure 44. Therefore, the ideal number of cartridge heaters would be 7 to 8 as that would heat up the fluid within 5 minutes. However, each cartridge heater requires its own external power supply so a new heating system might be considered in the next prototype.

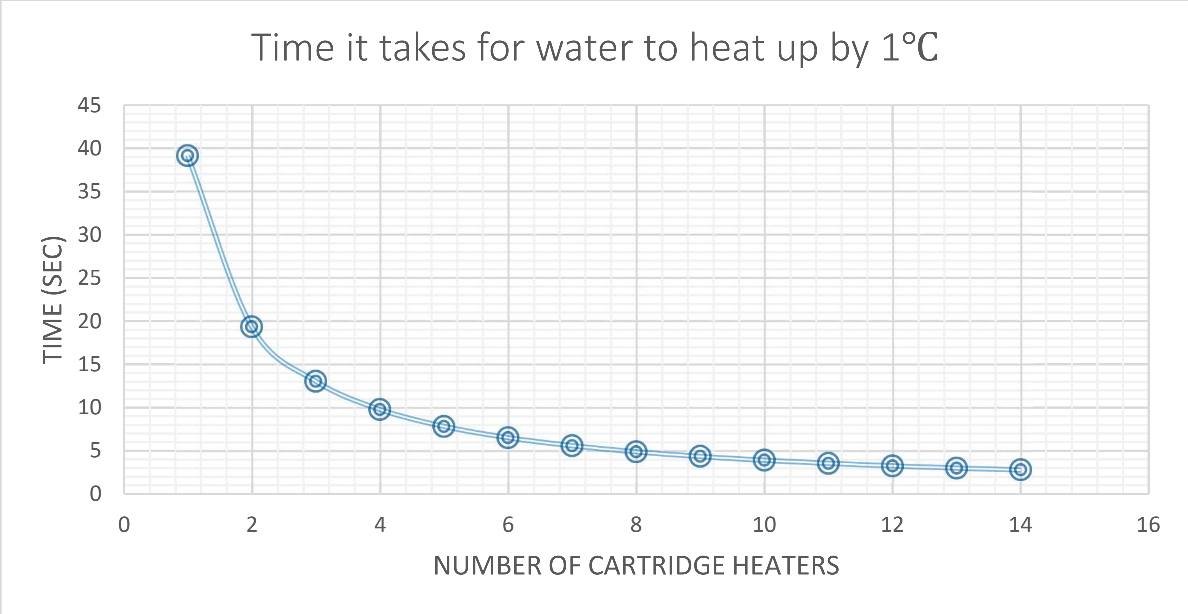


Figure 44:Graph showing how the number of cartridge heaters relates to how quickly it heats up 400ml of fluid by 1°C

To conclude, the strengths of the temperature control system built are that the electronics and programming function efficiently. The weakness is that the cartridge heaters chosen are not as powerful as they need to be as each one needs an external power source. However, this can also be avoided for short experiments by just putting in the right temperature fluid and using the cartridge heaters to maintain the temperature.

3. Pressure

In terms of creating the pressure, this machine is built to create pressures up to 120mmHg. These calculations assume laminar flow and an incompressible fluid. As the flow should stay mostly laminar, this is a fair assumption to make. However, due to the motion of the fluid and changes in diameter, it may not be wholly laminar. The fact that the equation assumes that the fluid is Newtonian is true for my experiment, where water was used, but may not always be the case as blood is a non-Newtonian fluid, which means that as shear-stress increases, viscosity decreases. As many experiments, however, would simulate the heart condition using thickened water, this is still a relevant way to increase the pressure for many experiments. The fact it is easily customisable also means that it should not be difficult to change it for a non-Newtonian fluid as well in the future, without changing the whole machine.

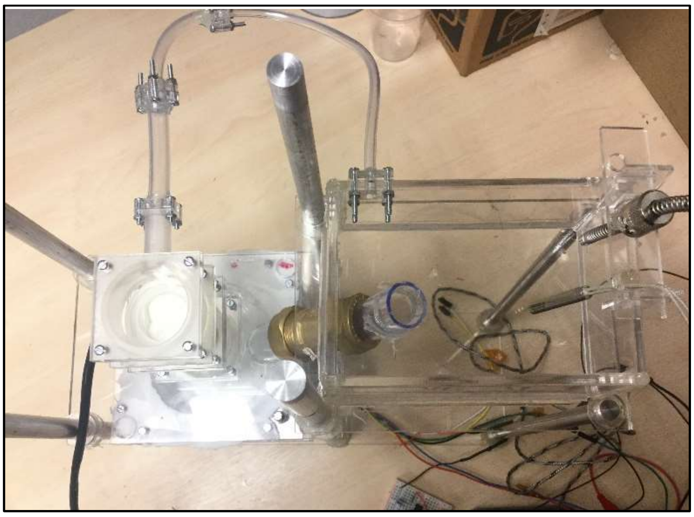


Figure 45:Picture of top of Bioreactor

4. LabVIEW Interface

The requirements for the LabVIEW interface would be that it was clear and easily understandable for any researcher to use. As can be seen from Figure 46, the interface can fit clearly onto one screen. There are clear instructions and the inputs are clearly described in the instructions. RPM, temperature and flow rate can all be monitored live throughout the experiment and exported afterwards. There is an additional programming aspect which is the Timer. This involves the researcher being able to input how much time they want the machine to run for in a standard time format and a progress bar, which shows how much time is left for the machine to run and how much time has passed. An emergency stop button will stop the whole machine in case of malfunction.

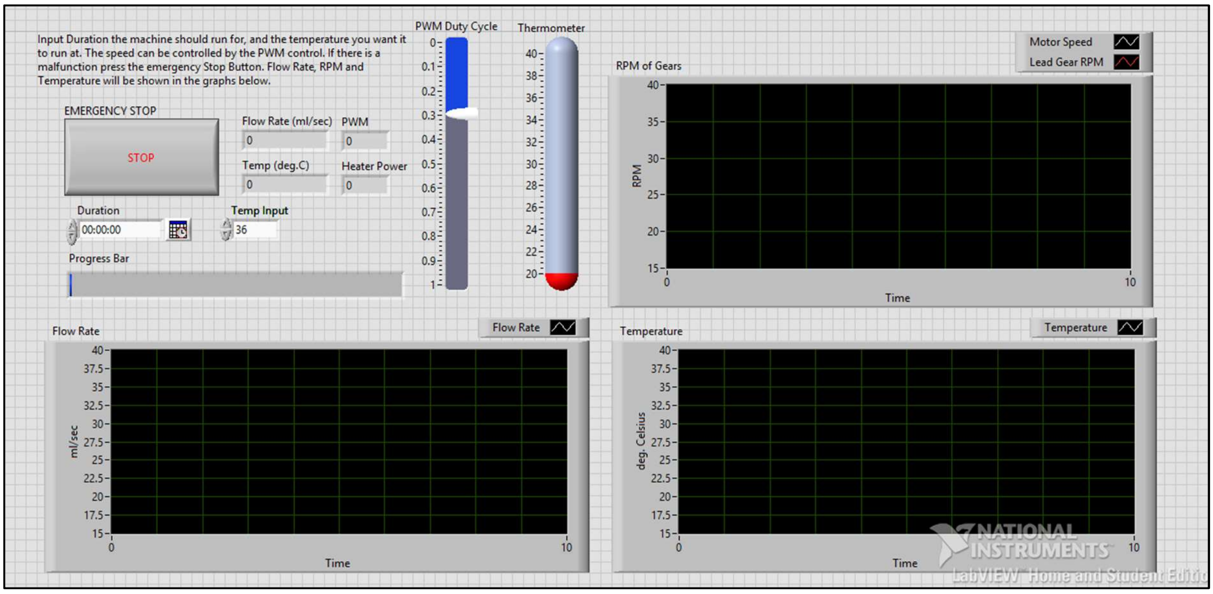


Figure 46:LabVIEW Interface

5. Viewing Window

After the design and manufacturing of the viewing window, a standard artificial heart valve was put in to compare visibility from the previous prototype (21). As you can see in Figure 47, there is a clear improvement in the view of the heart valve from one version to the other.

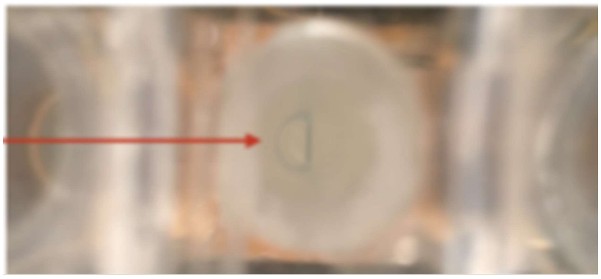
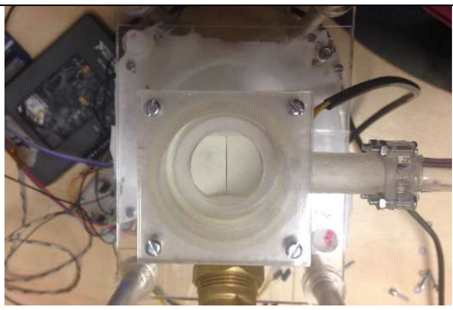
Previous Project	New Design
	

Figure 47:Photos showing the previous and new design comparison of the view of the heart valve (21)

6. Functionality of Machine

In terms of the functionality of the machine, the requirement was that it would be easy to dismantle so that it could be easily cleaned and customised depending on the researchers' needs. The total time to take out the heart valve and clean it or swap it is equivalent to the time it would take to undo 4 screws. The total number of screws needed to undo to clean the whole machine is 16, which is not much considering that before that was not possible due to the fact that it was glued together.

The other aspect is that the machine needs to move smoothly with the motor. This is demonstrated in the video attached on the memory stick.

Some glue was necessary to hold the top bellow platform down. However, the glue used was just glue gun glue which is not hard to undo and can easily be put back by just heating it up again.

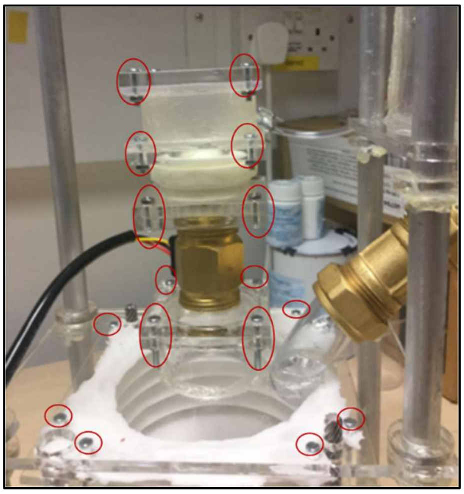


Figure 48:Picture identifying the screws that allow machine to be dismantled

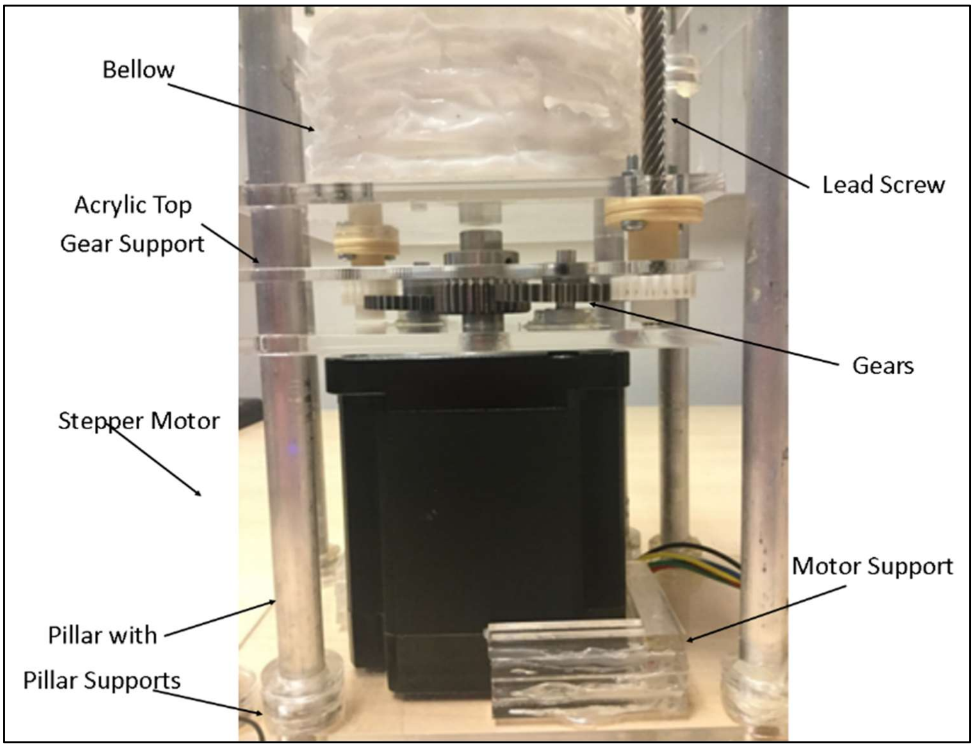


Figure 49:Picture of labelled base of machine

7. Cost

One of the main objectives of the project, especially considering the current competitors, was to make it competitively cheaper. This was achieved completely as all the components were bought with the budget of £400 given. The total cost including the parts that were recycled from the previous project brings the total cost to £927.17 but since the parts were cleverly recycled, this cost was £357.32. This can all be seen in the cost breakdown below;

Table 8:Breakdown of cost of project

COMPONENT	COST	COST (exc. Recycled)
Thermocouple	£23.20	£23.20
Pressure sensor	£13.56	£13.56
Cartridge Heater	£17.24	£17.24
Diligent motor adaptor	£62.80	£62.80
Stepper motor	£78.04	£78.04
Tubing	£19.58	£19.58
Silicon bellow	£26.08	£26.08
	£7.44	£7.44
Gear	£26.23	£26.23
Viewing Window		
Acrylic sheet	£26.97	£26.97
O-rings	£3	£3
Screws	£17	£17
Breadboard	£5.00	-
Jumper wires		0
MOSFETS & Op-Amp	£7.50	£7.50
MyRio	£455.00	-
Lead screws and nuts	£75.09	-
Flow Sensor	£10.26	-
Power Supply	£22.00	-
Pillars	£12.60	£12.60
Lead gears	£18.58	£18.58
TOTAL:	**£927.17**	**£357.32**

As can be seen from this, thanks to the support of the university and being able to recycle a few key components, the total savings to make this project was £569.85. Even without this, the cost would still be significantly less than the current models available on the market. This really shows the potential of this machine not to be outsourced and provide all the need to the researchers without having to use very expensive equipment.

8. Biocompatible

In terms of making the machine bio-compatible, the main change was introducing a silicone bellow instead of a latex one. This was conducted in a clean environment and silicone can be sterilised easily with alcohol. The other materials used were all plastic, so are very biocompatible. This was not too much of an issue in my project, as I would only be testing it with water, but was a good aim to allow for future development of the project.

4. Conclusions and recommendation for future

To conclude, the main requirements for the project were all met, as the final machine was designed and built from scratch with all components accounted for. The improvements on the last prototype included but are not limited to;

- Increased visibility of heart valve
- Temperature control
- Higher pressure control
- LabVIEW controlled programming for all components
- Modular design structure
- Pulsatile flow from the motor
- Increased Biocompatibility

In terms of future work for this machine, there is much scope for improvement and development.

- Bellow Manufacturing

The bellow manufacturing was successful when accounting the resources at hand. However, since it was developed in-house, it did not have a finish that would make it the most durable and water-tight component. This is vital when building a bellow pump so it would be worth outsourcing this, even if it costs upwards of £500 to make a durable pump system.

- Pressure Sensor

A further improvement could be implementing a pressure sensor into the LabVIEW interface which would monitor the live pressure between the heart valve and the tank. This would be very useful for data analysis later.

- Biochemical Heart Conditions

In this project I focused on the biomechanical heart conditions. However, a heart valve also is under many biochemical conditions. With the implementation of the temperature control, developing the chemical conditions would be supported by that development and would give the researchers further data to analyse and a better representation of what is happening in the calcification of the heart valve. These chemical conditions could vary from the pH and oxygen levels to implementing the calcium in the stream and monitoring these live.

- PCB[2] Electronics

All the electronics are currently exposed on a breadboard. However, this comes with issues such as wires getting detached easily and having to wire it up every time. An improvement

[2] Printed Circuit Board

would be turning the electronics onto a PCB which would allow researchers to easily wire up the machine, as well as leave it wired up permanently.

- Turning pump motion

In this model a simple straight pulsatile flow was explored to simulate the heart conditions. However, the heart also uses a twisting motion when pumping blood in the body creating some turbidity in the flow. This improvement to the machine could be implemented by changing the design of the bellow or the gearing but it would be an interesting direction in terms of the next prototype.

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Appendices

Appendix 1: Motor Comparison Chart

To choose the motor, the following comparison chart was made comparing the specifications of a variety of motors supplied by RS components. A list below describes what each of the criteria means.

Table 9: Comparison table of the different motors considered to replace the previous one

		Chosen Motor			Previously Used
NAME	As on website	**RS PRO Unipolar Bipolar, Unipolar Stepper Motor 1.8°, 4.31Nm, 6.4 V dc, 2 A, 6 Wires**	**RS PRO Unipolar Bipolar, Unipolar Stepper Motor 1.8°, 6.47Nm, 7.6 V dc, 2 A, 6 Wires**	**RS PRO Unipolar Bipolar, Unipolar Stepper Motor 1.8°, 1.32Nm, 8.6 V dc, 1 A, 8 Wires**	**Sanyo Denki Bipolar Hybrid Stepper Motor 1.8°, 1.6Nm, 24 V dc, 2 A, 4 Wires**
Available now	In stock	Y	Y	N (5/2/2019)	Y
Price		**£78.04**	**£211.51**	**£83.02**	£69.00
Compatible with Digilent motor adaptor	Volatge between 6-16V	Y	Y	Y	N
Step Angle (deg.)	Precision of motor	1.8°	1.8°	1.8°	1.8
Holding Torque (Nm)	The force it can spin at	4.31Nm	6.47Nm	1.32Nm	1.6
Number of Wires		6	6	8	4
Voltage Rating	Voltage input required	6.4 V dc	7.6 V dc	8.6 V dc	24 V dc
Frame Size	Dimensions of motor	85.8 x 85.8mm	85.8 x 85.8mm	56.4 x 56.4mm	56 x 56mm

Stepper Motor Type	Bipolar is compatible with Motor adaptor	Bipolar, Unipolar	Bipolar, Unipolar	Bipolar, Unipolar	Hybrid
Shaft Diameter		14mm	14mm	6.35mm	6.35mm
Current Rating		2 A	2 A	1 A	2 A
Winding Arrangement		Unipolar	Unipolar	Unipolar	Bipolar
Shaft Length		167mm	197mm	114.1mm	19.1mm
Depth		96mm	126mm	77.5mm	77.3mm
Resistance Per Phase		3.2Ω	3.8Ω	8.6Ω	2Ω
Standards Met		RoHS Compliant	RoHS Compliant	RoHS Compliant	

Appendix 2: Finding the windings of the Stepper Motor

To connect the motor to the adaptor, I had to find out which wires belonged to which coils. This was made more complicated by the fact that it was a hybrid motor, so had two wire taps used only during the unipolar wiring. If there was a resistance between the wires, then they were part of the same coil; if there wasn't any, they were not. Once that was found out, the next step was working out which wires were the coil ends and which were the taps. Between a tap and coil end, the resistance was half that found between both coil ends. The results of the findings can be seen in Table 10, and a diagram of the final colours and windings can also be seen in Figure 12 in the report.

Table 10: Table showing the resistances between the motor wirings and image to represent the coil ends and wires with the respective colours found.

Resistance (Ω)	Red	Blue	Green	Yellow	Black	White
Red		7.3	0	0	0	4.1
Blue			0	0	0	4.1
Green				4.1	7.3	0
Yellow					4.0	0
Black						0
White						

Appendix 3: Motor Gear Comparison

This table shows the comparison between the gear bought for the new motor chosen and the comparison with the previous gear used.

Table 11: Table showing Comparison of Motor Gears to show changes between previous project and current one used.

	RS PRO Steel 20 Teeth Spur Gear, 20mm Pitch Diam. , 16mm Hub Diam. , 8mm Bore Diam.	**RS PRO Stainless Steel 30 Teeth Spur Gear, 30mm Pitch Diam. , 25mm Hub Diam. , 8 → 16mm Bore Diam.**
	PREVIOUS GEAR	NEW GEAR
	£9.86	**£26.23**
Module	1	1
Number of Teeth	20	30
Material	Steel	Stainless Steel
Bore Diameter	8mm	8 → 16mm
Hub Diameter	16mm	25mm
Pitch Diameter	20mm	30mm
Face Width	10mm	10mm

Appendix 4: Hall Effect Sensor

The flow sensor (G3/4" Hall Effect Liquid flow Sensor Switch Flowmeter) is being recycled from the previous project. This flow sensor is based on a hall effect. The hall effect is the principle that when a magnet is brought close to a conducting plate, the electrons are deflected, causing a change in voltage to occur. In our case, as shown in Figure 50, the wheel which is pushed by passing water, changes the hall sensor's magnetic field. This means that the magnet is either pushed forwards or backwards at different rates creating a wave of voltage in the circuit (30). Used in conjunction with an op-amp or programming, this can create a very clear digital output which is more easily manipulated when it comes to measuring flow rate.

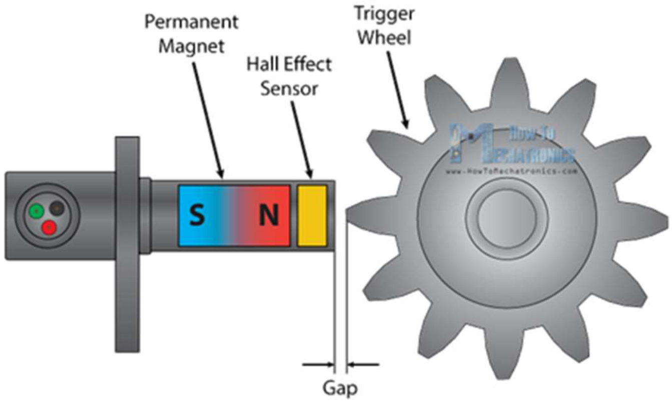

Figure 50:Diagram of hall effect flow sensor (30)

Appendix 5: Pressure Adjustments Spreadsheet

Also attached on the memory attached.

	SUMMARY		
Input desired Pressure	PRESSURE	120	*mmHg*
Input Fluid Viscosity	VISCOSITY	0.00089	*Pas*
Total tubing length	TOTAL TUBING LENGTH	0.311	*m*
3mm Tubing Length for Experiment	3mm TUBING LENGTH	0.161	*m*
Price to buy 3mm Tube	PRICE	£ 0.14	

Pressure Difference Between pump and water chamber	15998.68421	Pa	120	mmHg					
Average Flow rate Provided by pump	0.000209	m^3/s							
Total Resistance	76548728.28	Pas/m^3							
Blood Viscosity	0.00089	Pas							
	Diameter (m)	**Length (m)**	**Number**	**Resistance**		**Tubing Length to buy**	**COST PER METER**	**COST**	
Initial Tube	0.036	0.030	1	648					
Sensor	0.019	0.044	1	12243					
Valve	0.030	0.025	1	1119					
Viewing window	0.040	0.020	1	283					
Tube 1	0.010	0.030	1	108786		0.030			
Tubes 2	0.008	0.050	1	442650		0.050	£ 1.23	£	0.06
Tubes 3	0.005	0.070	1	4061329		0.070	£ 1.11	£	0.08
				4627057					
			Total Resistance Left:	**71921671**					
Tube 4	0.003	0.161	1	71921671		0.161	£ 0.87	£	0.14
	TOTAL LENGTH:	**0.311**						**£**	**0.28**

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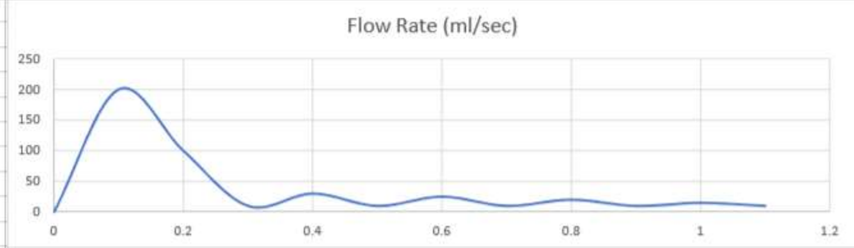
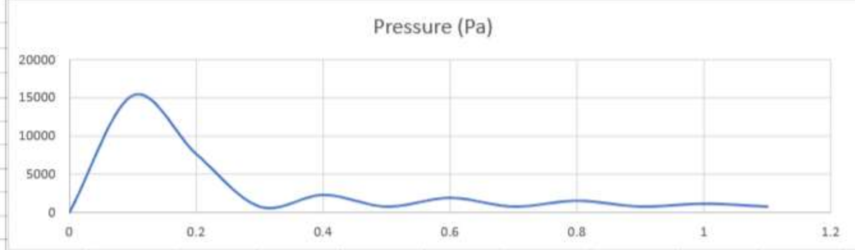
Pressure Difference Between pump and water chamber	=D1*(101325/760)	Pa	=SUMMARY!C4	mmHg				
Average Flow rate Provided by pump	0.000209	m^3/s						
Total Resistance	=B1/B2	Pas/m^3						
Blood Viscosity	=SUMMARY!C5	Pas						
	Diameter (m)	**Length (m)**	**Number**	**Resistance**		**Tubing Length to buy**	**COST PER METER**	**COST**
Initial Tube	0.036	0.03	1	=(128*B6*C9)/((3.14159265)*(B9^4))				
Sensor	0.019	0.044	1	=(128*B6*C10)/((3.14159265)*(B10^4))				
Valve	0.03	0.025	1	=(128*B6*C11)/((3.14159265)*(B11^4))				
Viewing window	0.04	0.02	1	=(128*B6*C12)/((3.14159265)*(B12^4))				
Tube 1	0.01	0.03	1	=((128*B6*C13)/((3.14159265)*(B13^4))/D13)		=C13*D13		
Tubes 2	0.008	0.05	1	=((128*B6*C14)/((3.14159265)*(B14^4))/D14)		=C14*D14	1.23	=H14*G14
Tubes 3	0.005	0.07	1	=((128*B6*C15)/((3.14159265)*(B15^4))/D15)		=C15*D15	1.11	=H15*G15
				=SUM(E9:E15)				
			Total Resistance Left:	**=B4-E16**				
Tube 4	0.003	=(E18*(3.14159265)*(B18^4))/(128*B6)	1	=(D18*E17)		=C18*D18	0.87	=H18*G18
	TOTAL LENGTH:	**=SUM(C13:C18)**						**=SUM(I14:I18)**

	Flow Rate		Resistance	
Time (sec)	(ml/sec)	m^3/sec		Pressure
0	0	0	76548728.28	0
0.1	200	0.0002	76548728.28	15309.75
0.2	100	0.0001	76548728.28	7654.873
0.3	10	0.00001	76548728.28	765.4873
0.4	30	0.00003	76548728.28	2296.462
0.5	10	0.00001	76548728.28	765.4873
0.6	25	0.000025	76548728.28	1913.718
0.7	10	0.00001	76548728.28	765.4873
0.8	20	0.00002	76548728.28	1530.975
0.9	10	0.00001	76548728.28	765.4873
1	15	0.000015	76548728.28	1148.231
1.1	10	0.00001	76548728.28	765.4873

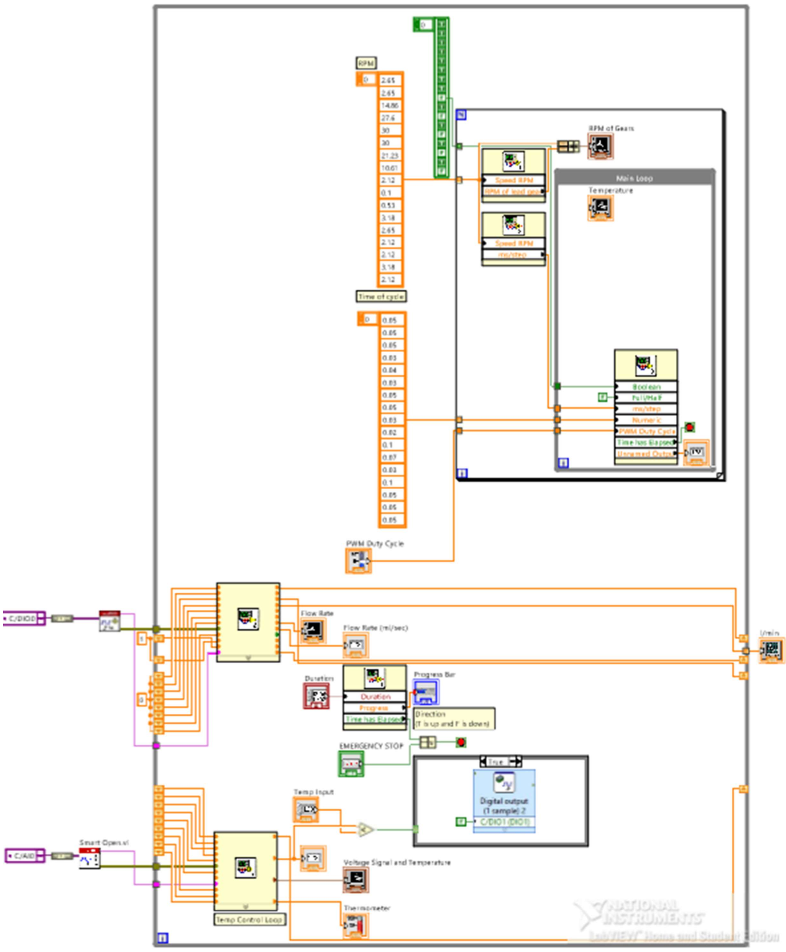


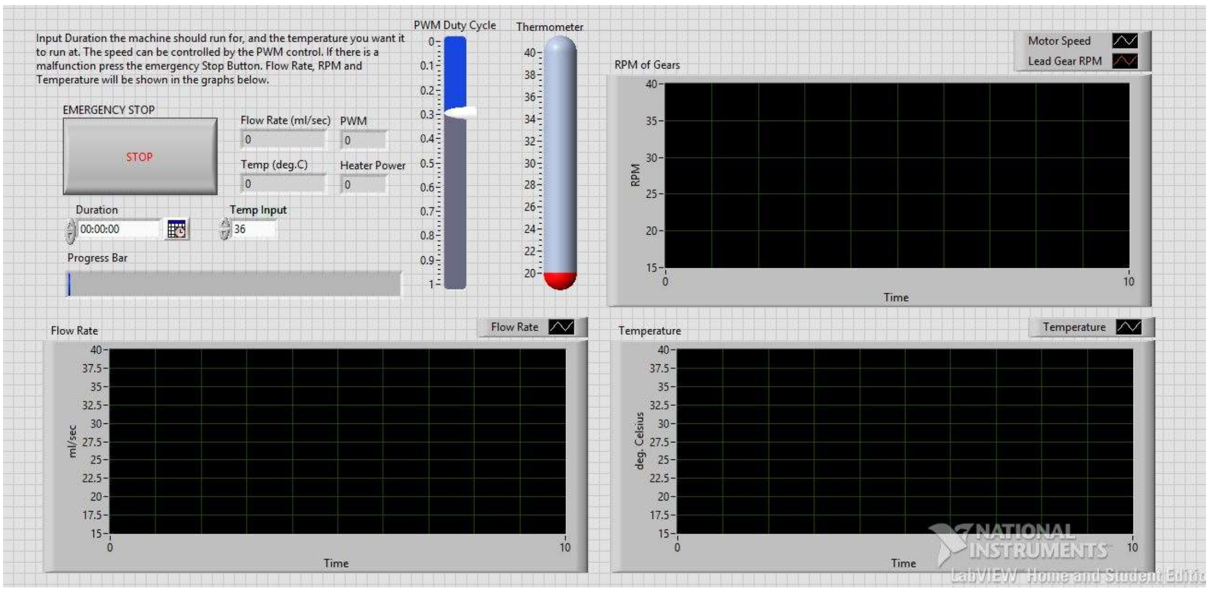


Appendix 6: LabVIEW Program

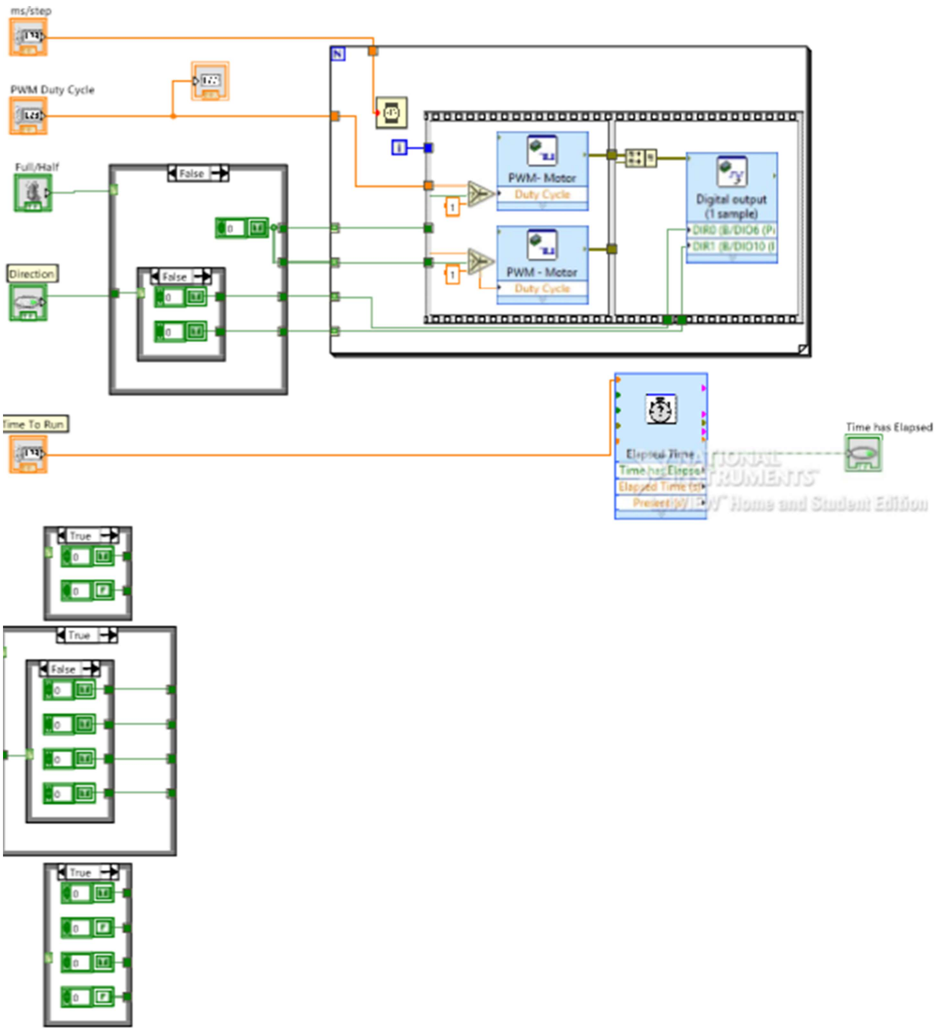
This whole program is attached on the memory stick which is given with this document.

Main Program



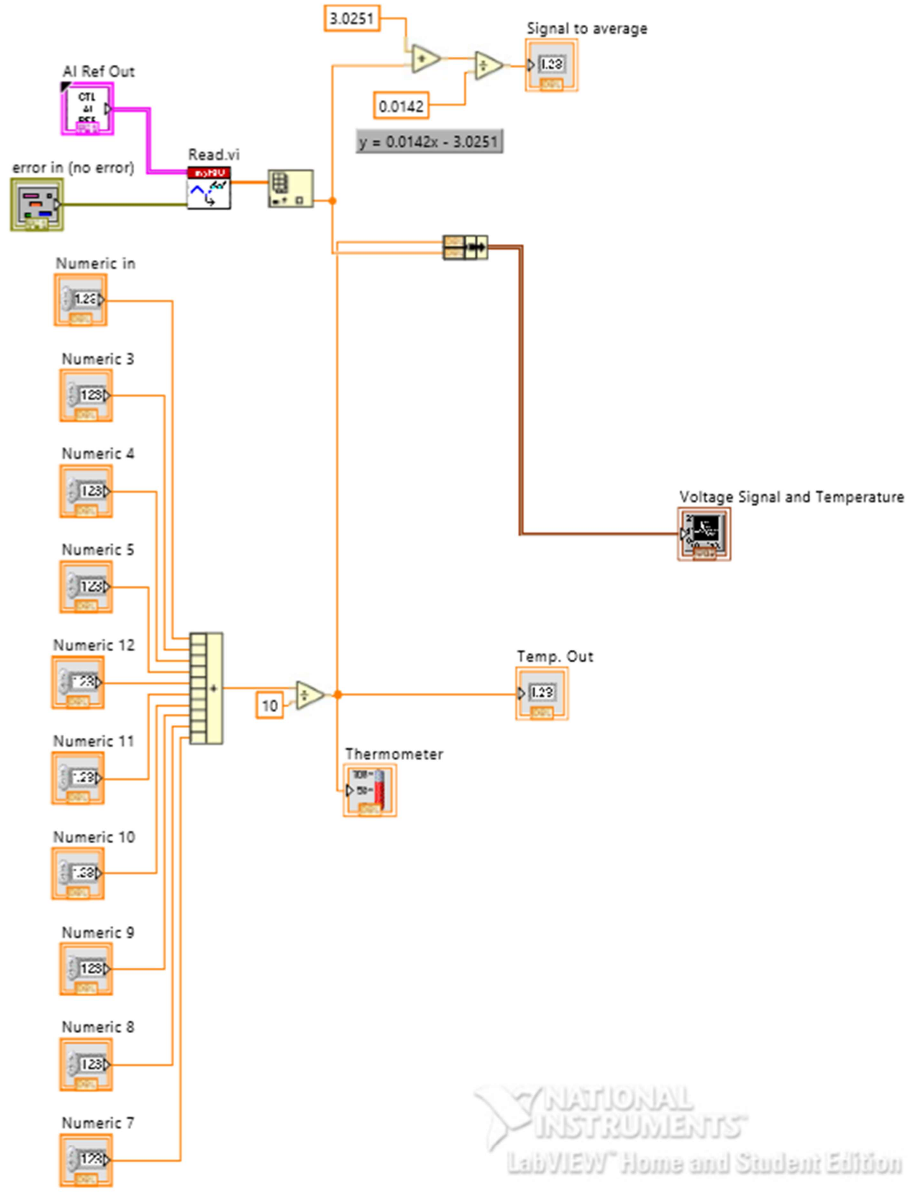


Motor Turning Program



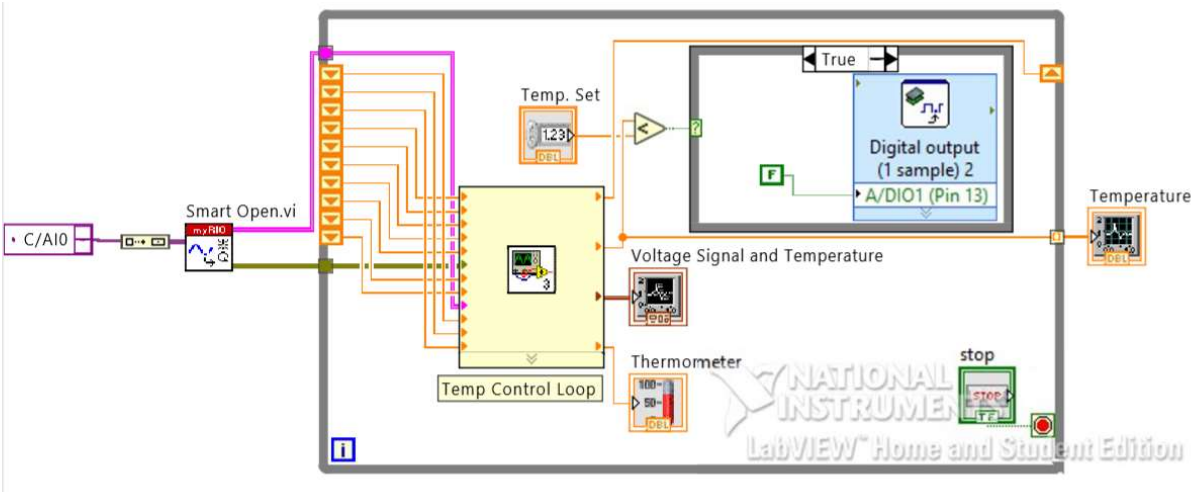
RPM to ms/step conversion

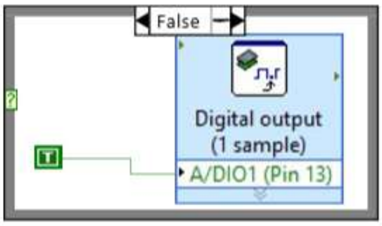
Thermocouple Conversion to Temperature



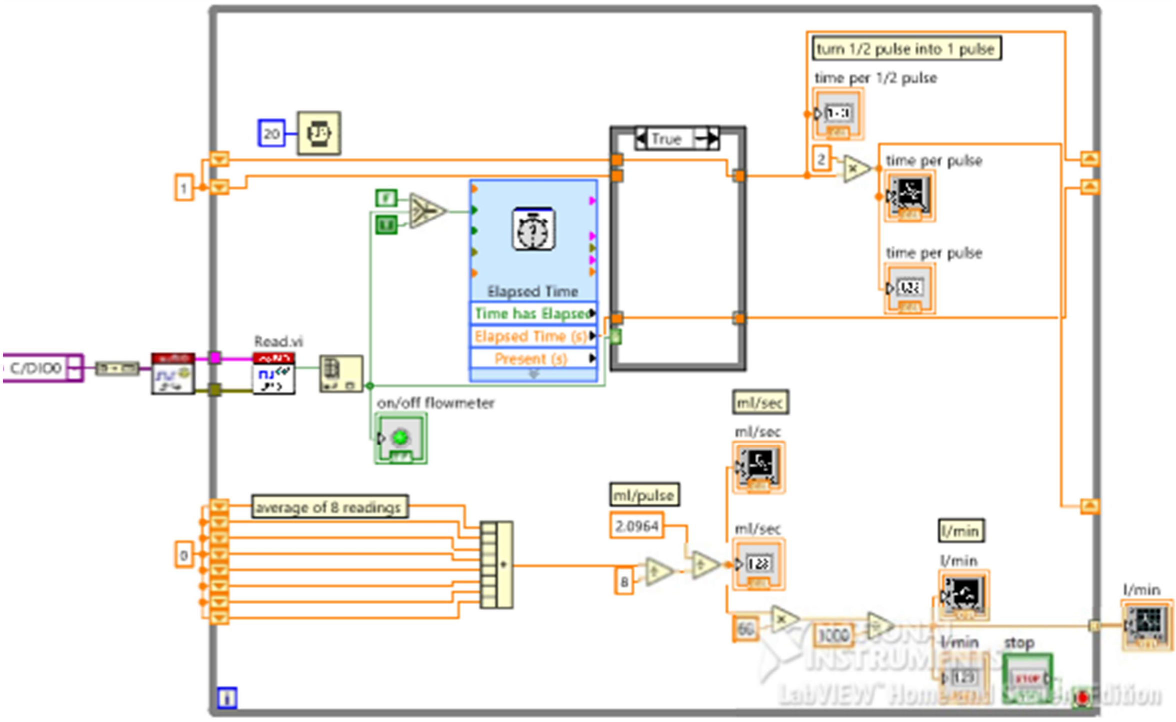
RPM to RPM of Lead Gear conversion

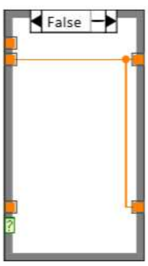
Temperature Control Program



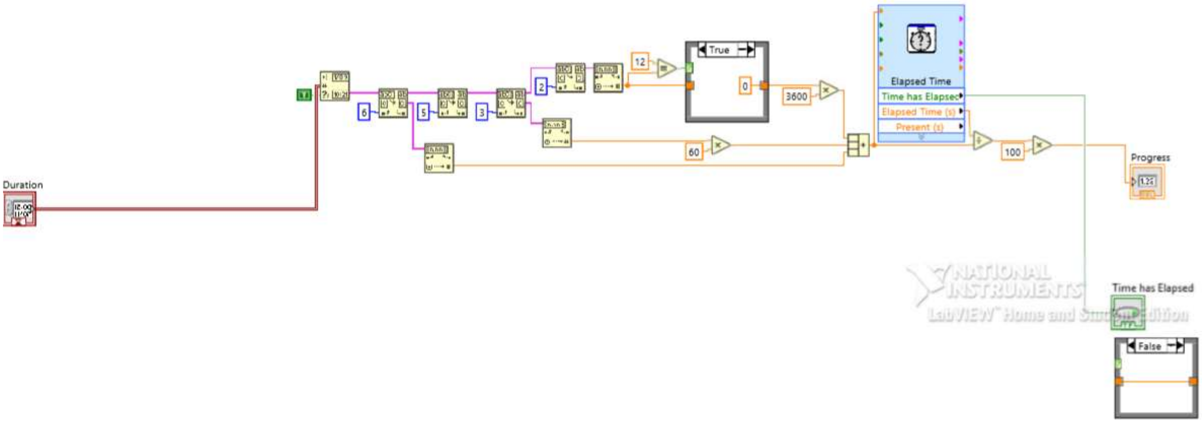


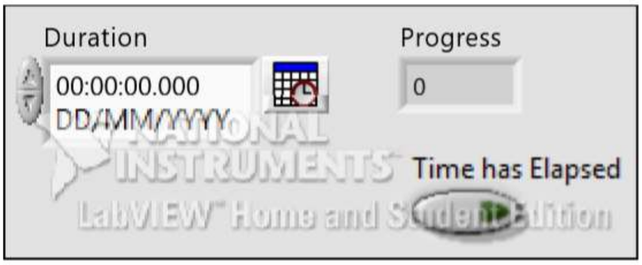
Flow Sensor Program





Timer of Program





Appendix 7: CAD Drawings

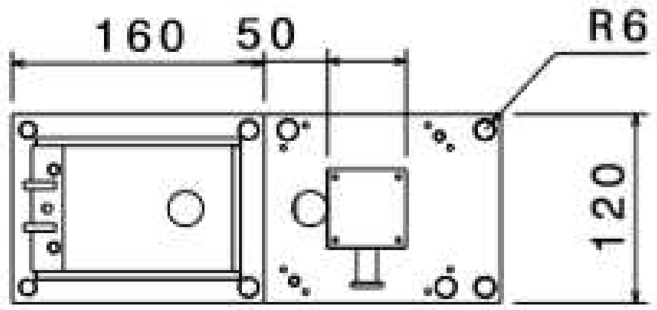
Technical Drawing of Bioreactor to Test Heart Valves

SCALE: 1:4

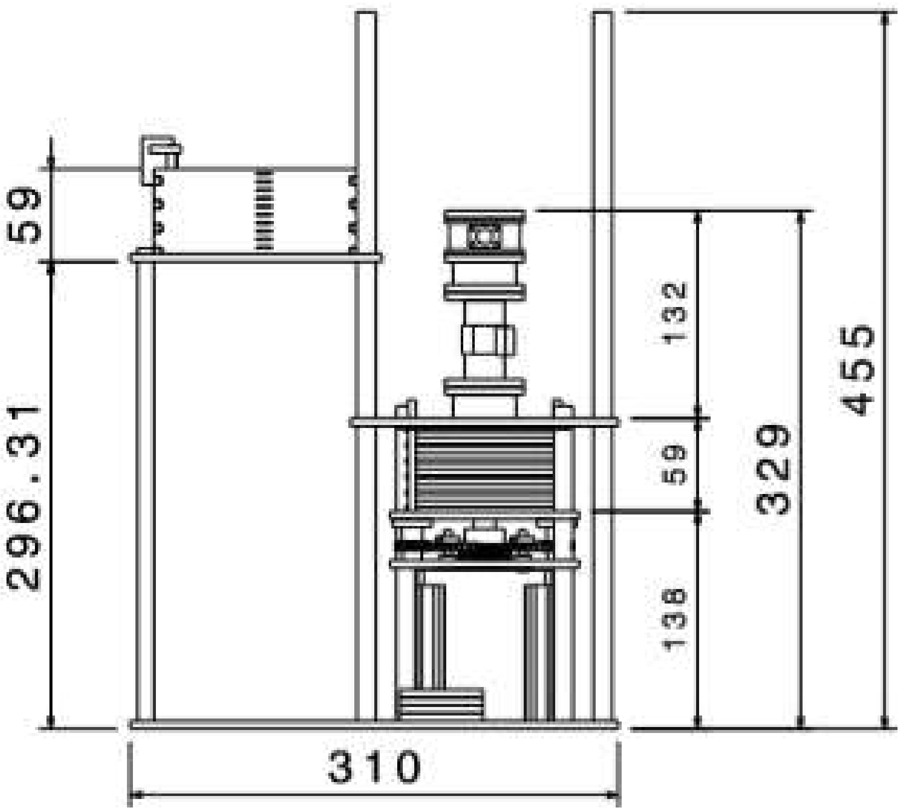
Third Angle Standard

MEASUREMENT UNITS: mm

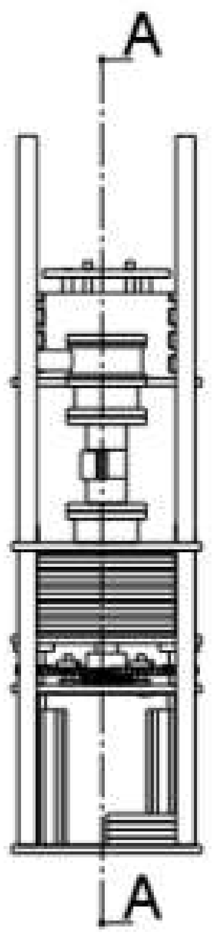
AUTHOR: Julia Elkouby



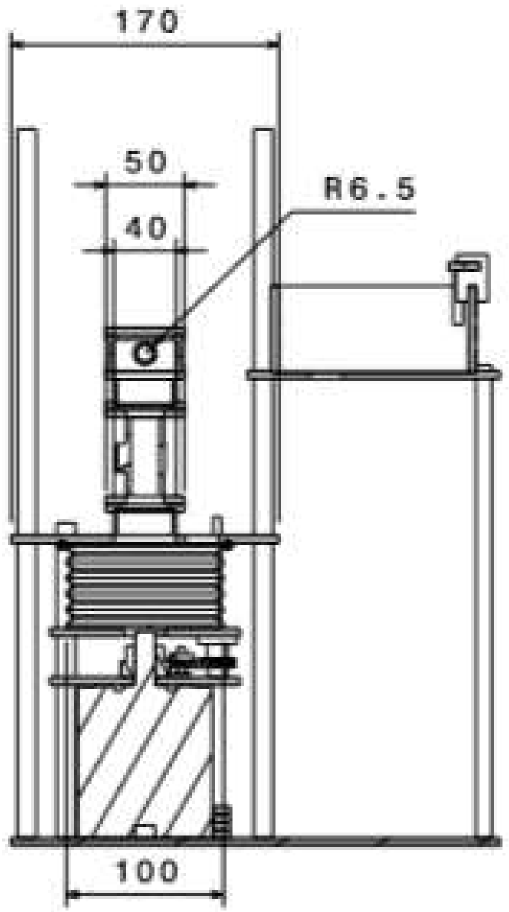
Top view



Front view



Right view



Section view A-A

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