

Design Optimization of Oxygenated Fluid Pump

by

Bruno Aiala Piazzarolo

Submitted to the
Department of Mechanical Engineering
in Partial Fulfillment of the Requirements for the Degree of

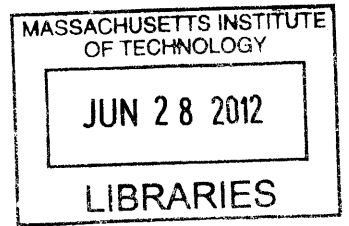
Bachelor of Science in Mechanical Engineering

at the

Massachusetts Institute of Technology

June 2012

ARCHIVES



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ABSTRACT

In medical emergencies, an oxygen-starved brain quickly suffers irreparable damage. In many cases, patients who stop breathing can be resuscitated but suffer from brain damage. Dr. John Kheir from Boston Children's Hospital created a compressible fluid that can re-oxygenate blood quickly in patients with asphyxia and cardiac arrest. Because the fluid is compressible, the set infusion rate on an ordinary pump does not necessarily indicate what is delivered. In addition, the fluid is provided at a 90% gas by volume concentration and is extremely viscous.

The goal of this project is to create a pump to deliver a specified volumetric flow rate of the oxygenated fluid created by the doctor. The pump design uses a bellows with force feedback calibration to pump 1 liter of fluid over 5 minutes and mix the concentrated 90% form with saline without degradation to form a 70% concentrated form with the viscosity similar to that of blood. My part in the project was to create the control system that would drive the pump using a force feedback and to optimize the design of the pump

The oxygenated fluid pump built can successfully store and dispense one liter of fluid, mix the concentrated form of the oxygenated fluid with saline, maintain sterility, and preserve the fluid's properties, all in a cost appropriate manner. It is a modular design that can easily be modified to improve its performance. Further testing is required to tune the control system and ensure that the flow rate is accurate to $\pm 10\%$. The pump is mostly being used as a research tool in order to run tests that will help characterize the fluid and later can be used for small and large animal testing.

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ACKNOWLEDGEMENTS

I would first and foremost like to thank Professor Alex Slocum and teaching assistants/mentors Nevan Hanumara and Nikolai Begg for their expert advice, support and guidance in this project. I would also like to thank our clinician Dr. John N. Kheir, M.D for providing us with an exciting project that has tremendous potential to save lives.

I would also like to thank partners Alex Mason and Kristen Pena. We formed a team of extremely dedicated scholars determined to find a solution to the problem presented by our clinician. Without their support and wisdom this project would not have been successful.

Last but not least I would like to thank everyone who helped in the project in one way or another, mainly Lesley Yu (National Instruments representative for MIT) who provided great technical skills and support. She provided help with Labview to create the backbone of the control system used in this project. Thanks to Blow Molded Specialties, the Center for Bits and Atoms, 3-D Printsmith, and the MIT Hobby Shop for providing support with special parts and manufacturing needed for this project.

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INTRODUCTION

1.1 PURPOSE

The purpose of this project is to explore in detail the design and optimization of a pump to deliver a specified volumetric flow rate of an oxygenated fluid. Dr. John Kheir works at children's hospital and has developed a fluid that can re-oxygenate blood in non-breathing patients. The fluid is compressible and presents many problems when being pumped using conventional medical devices such as peristaltic or syringe pumps. The main problem with these pumps is that since the fluid is compressible, the set infusion rate does not necessarily indicate how much fluid is dispensed.

This thesis will investigate how a pump was developed, including the analysis performed to correctly choose and size different components such as motors and sensors. It will detail how the pump was designed to be modular, low cost, and have the potential to save life.

1.2 MOTIVATION

In medical emergencies, an oxygen-starved brain quickly suffers irreparable damage. In many cases, patients who stop breathing can be resuscitated but suffer from brain damage. The

objective is to build a device to infuse a specialized suspension which oxygenates the bloodstream, allowing for more time to resuscitate a patient before they suffer brain and/or organ damage. The oxygenated fluid and delivery system could improve medical outcomes for patients with cardiac arrest, which currently has a neurologically intact survival rate of sub-five percent.

Oxygen is required to maintain cell homeostasis in the human body. Serious organ injury and death can occur in critically ill patients experiencing asphyxia, lung injury, and cardiac arrest because they cannot consume the amount of oxygen needed to survive. This oxygenated fluid raises oxyhemoglobin saturation upon injection, and can improve the survival rates of critically ill patients [1]. Because current medical infusion pumps make the assumption that the amount that the plunger depresses is directly proportional to the volume injected, the injected volume does not to be measured. However, when a fluid is compressible, the amount dispensed by the plunger being depressed is not proportional to the volume injected. Therefore, the amount of oxygen a patient receives from the oxygenated fluid is not known. The device outlined in this thesis will account for this discrepancy.

1.3 PUMP OVERVIEW

The pump design can be broken up into three modules: the oxygenated fluid dispenser, the saline dispenser, and the mixer. The oxygenated fluid dispenser consists of the bellows, a linear actuator to compress the bellows, and a feedback control system. The saline dispenser consists of a peristaltic pump. A plate attached to linear actuator will compress the bellows. Between the plate and the bellows top is a load cell that will provide the data for the feedback control. Both the oxygenated fluid dispenser and the saline dispenser have tubing that guides the respective

fluids to a wye junction where they meet and are then mixed by a static helical mixer. The system diagram is shown below in Figure (1.1).

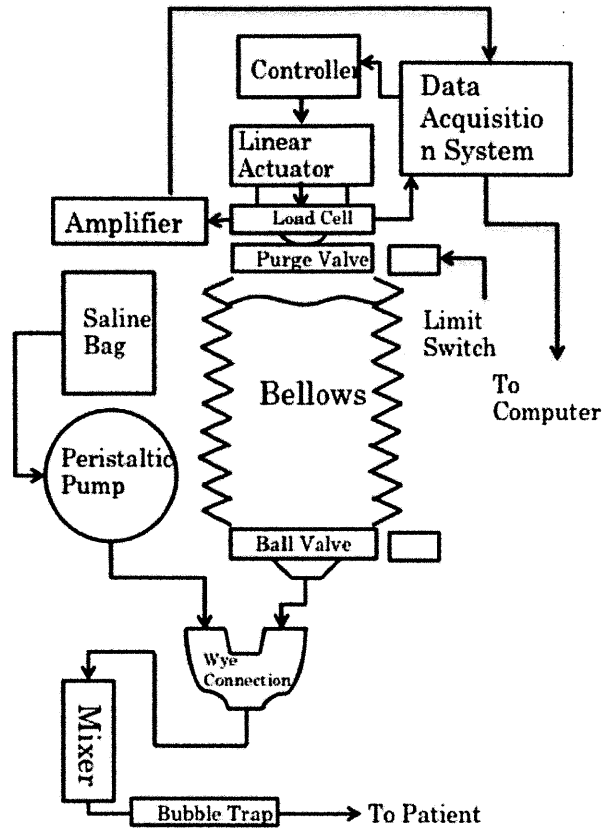


Figure (1.1) – This figure shows the overall system diagram.

The flow also passes through a bubble trap before the fluid can be injected into a patient. More detail about each of the major components, how they were selected and their functions is outlined in Chapter 3.

1.4 PRIOR ART

Research was done to find previous pumps that were used for other unconventional fluids. From exploring these different options, a few ideas were incorporated into the design for the oxygenated fluid pump described in this paper.

1.4.1 Belmont Rapid Infuser

The Belmont Rapid Infuser is a high efficiency electric warmer coupled with a high-speed peristaltic pump [2]. It is capable of measuring flow rate, the temperature at the inflow and outflow of the heater, patient temperature, and line pressure. It can also detect and trap air bubbles before infusion into the patient. This device however uses a peristaltic pump, which is not sufficient for the oxygen suspension because the pump causes cavitation in the fluid. It is also not capable of compensating for fluid compression when calculating flow rate. Though the full system cannot be used, the bubble trap was extracted for use in our design because it should be capable of handling high flow rates.

1.4.2 Enhanced ACL Repair Gun

The enhanced ACL repair gun is the product of a previous 2.75 project that allows for proper mixing and heating of a fluid [3]. The gun has a heating unit on one end of gun and an adjustable auger that mixes fluid when deployed; this product faced the mixing challenge that also needs to be addressed in the dilution of the oxygenated fluid. The auger inspired the use of passive mixing techniques involving helical structures for this device.

FUNCTIONAL REQUIREMENTS

2.1 DESIGN REQUIREMENTS

The primary goal of this design was to deliver 1L of the oxygenated fluid at a specified volumetric flow rate. From this and in discussion with Dr. Kheir the following functional requirements and accompanying parameters were identified:

Table (2.1) – The functional requirements of the Oxygenated Fluid Pump.

Functional Requirements	Design Parameters
Flow rate accurate to $\pm 10\%$ <i>(must deliver fluid between 10 and 200mL/min into 14 gauge catheter)</i>	Flow sensors, valves + controller <i>(“the set infusion rate on the pump does not indicate what is delivered”)</i>
Can handle at least 1L fluid	Reservoir system
Mix with saline	Static or kinetic mixers
Will not degrade fluid	Limits on pressure <i>(upper limit: 10psi)</i>
Maintains sterility	Limit contact with fluid
Portable <i>(nice to have)</i>	Minimize size, mass
Maintainable or disposable	Cheap plastic parts
Cost appropriate	
Traps bubbles	Include bubble trap

The functional requirements in Table 2.1 were the minimum requirements that had to be met to complete the task Dr. Kheir gave to us and to ensure we had a reliable pump. They are outlined in this chapter with more detail, including how some of the decisions for the design parameters were made.

2.1.1 Cost

Medical devices are extremely expensive. Standard syringe pumps and peristaltic pumps cost anywhere from \$500-\$4000. The goal for the oxygenated fluid pump described in this project was to keep it around this same price range. A table of all the components purchased and the total cost can be found in Appendix (B).

2.1.2 Flow Rate

The flow rate for our pump is relatively high, set by Dr. Kheir. He determined that it would take about 1L of the oxygenated fluid for a human to survive around 5 minutes. Therefore, the flow rate was established to be 1L in 5 minutes or 200mL/min or around 3.3mL/s with an accuracy of $\pm 10\%$. This is how much fluid needed to be pumped into the patient at a 70% concentration. There are separate flow rates however for the peristaltic pump and the bellows pump. Since the fluid started at a concentration of 90% and needed to be mixed to 70%, it was determined that a mixing of about 250mL Saline and 750mL of the 90% concentrated oxygenated fluid was necessary. Therefore, the flow rate for the peristaltic pump was set at 250mL in 5 minutes or 50mL/min or about 0.83mL/s. The flow rate for the bellows pump was determined to be 750mL in 5 minutes or 150mL/min or 2.5mL/s.

2.1.3 *Mixing*

As mentioned before the oxygenated fluid needs to be mixed from a 90% concentration to a 70% concentration. A static mixer was chosen instead of a kinetic mixer because a kinetic mixer could potentially damage the bubbles. Also, it wasn't too difficult to mix the oxygenated fluid with saline and a static helical mixer was determined to be enough. To mix the fluid with saline there were two different paths that met at a wye junction and then into the helical mixer.

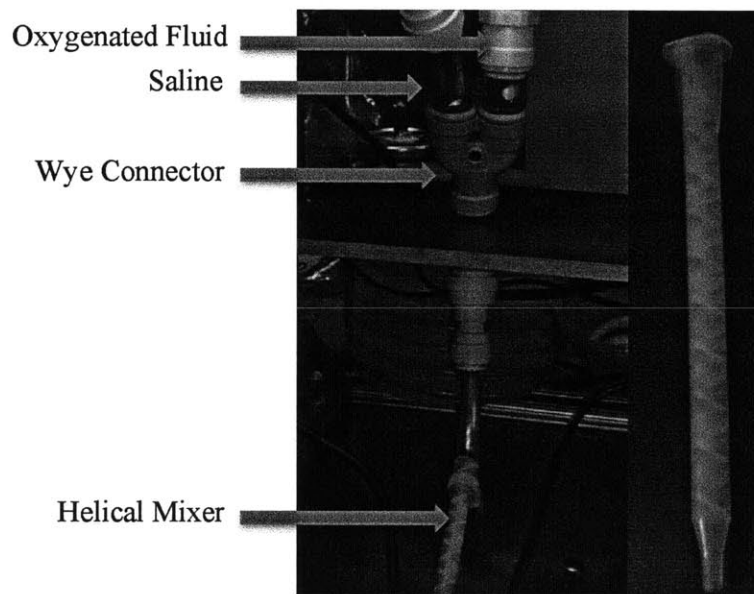


Figure (2.1) – Layout of fluid path into the helical mixer and close up view of mixer.

The helical mixer used was from an epoxy gun. From running some basic tests it was determined that this mixer did an adequate job at mixing the fluid. More information of mixing and determining the quality of the fluid after mixing can be found in section 6.1

2.1.4 Capacity

Dr. John Kheir determined that an appropriate amount of fluid to be pumped into a human in order to ensure proper oxygenation was 1L. Syringes that hold 1L of fluid are hardly found in the medical field and the frictional forces found in a syringe are hard to deal with. Taking into account that this had to be a large compressible container, a bellows was found to be appropriate. Bellows have predictable forces that can be easily modeled as a spring and are found in a variety of sizes. More information on bellows and volume can be found in section 3.2.4.

2.1.5 Medical-Specific Requirements

Not degrading the fluid, maintaining sterility, trapping bubbles and the accuracy of the flow rate are all medical-specific requirements. It is extremely important that during the pumping process, no more than 10 psi is exerted on the fluid in order to ensure the fluid is not damaged. This requirement was also given to us by Dr. John Kheir. It is important to maintain sterility throughout the pumping process. Therefore, the pump was design so that the major components are replaceable. For example, the mixer is cheap and replaceable, the bellows pump functions as a cartridge in which you can just replace the bellows with another one once empty, and the saline pump only requires the saline bag to be changed. Trapping the bubbles is an extremely important part of the design because if there are large bubbles flowing into the patient it can cause an embolism. Last but not least the accuracy of the flow rate is extremely important requirement because the exact amount of fluid being pumped into a patient needs to be known. The whole pump is designed to fulfill this functional requirement first and foremost.

3.1 SYSTEM FLUID MODEL

Sizing of different components in designing the Oxygenated Fluid Pump required correctly modeling the fluid as it moved through the pump. Because of the fluid viscosity, it is important to size the components correctly to achieve the desired flow rate. The modeling done in this section helped size the bellows and select the motor and load cell.

3.1.1 Pressure

Basic pressure analysis was done using the Hagen-Poiseuille equation:

$$\Delta P = \frac{8\mu L Q}{\pi r^4} \quad (3.1)$$

where:

ΔP is the pressure drop

L is the length of the pipe

Q is the volumetric flow rate

r is the radius of the pipe

μ is the viscosity of the fluid

This equation is for finding the pressure drop in a pipe where the length is much longer than the diameter of the pipe, which is true in this case. However, this equation also assumes that the flow is laminar, viscous and incompressible. The fluid being pumped in this case will meet all of these

assumptions except that it is compressible. This analysis is still valid however because its purpose is to make a first order approximation that will allow for the correct sizing of components.

Standard tubing was used, along with luer locks (medical grade connectors) to make the initial layout of our device shown in Figure (3.1).

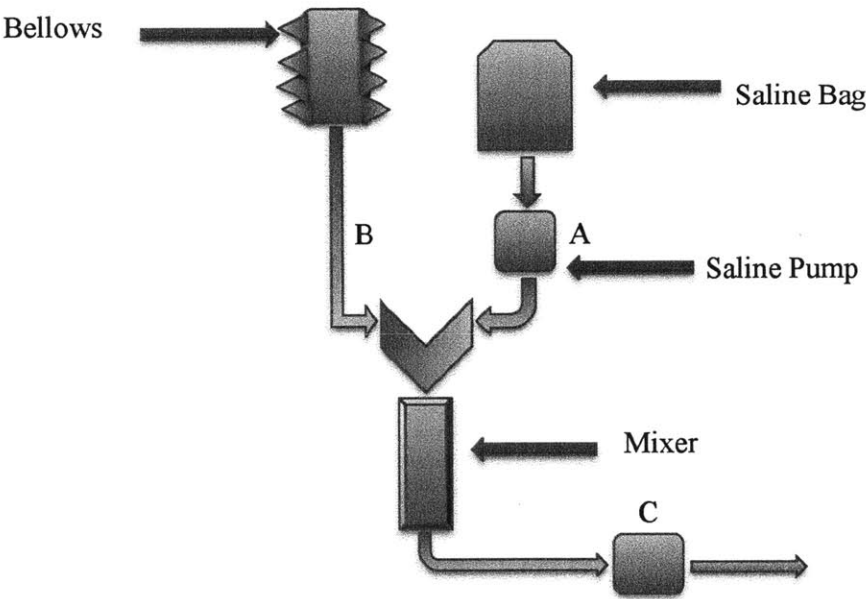


Figure (3.1) – This figure shows the basic layout of the pump in order to fulfill the functional requirements.

Sections A, B and C represent the different sections of tubing that fluid flows through. Using this general layout and the Poiseuille equation (3.1), other dimensions were approximated (such as radii and lengths of tubing) and analysis was done to approximate the resistance from each section of tubing, shown in Table (3.1).

Table (3.1) – This table shows the analysis done using Poiseuille’s equation.

	Section		
	A	B	C
Length (m)	0.7112	0.3556	1.8669
Radius (m)	0.003175	0.003175	0.002
Q (m ³ /sec)	8.33333E-07	2.5E-06	3.33333E-06
μ (Pa*sec)	0.001	0.001	0.001
P_final (Pa)	990	990	-
ΔP	14.85	22	990
P_initial (Pa)	1005.27	1,013	990
ΔP (psi)	0.15	0.15	0.14

The pressures calculated are gauge pressures. The flow rates in each section come from the functional requirements of the pump. Because fluid needs to be mixed from 90% gas by volume to 70% gas by volume and the desired flow rate is 1L in 5 minutes, it was possible to calculate the desired flow rates for sections A,B and C. By mixing 150mL/min (in section B) of the oxygenated fluid at a 90% gas by volume concentration with 50mL/min (in section A) of saline, the final concentration would be at 70% gas by volume at a flow rate of 200mL/min (in section C). These flow rates were then converted to SI units and were used in the calculations in Table (3.1). From this analysis we were able to get a basic idea of the pressure drops in each section, however, this initial analysis was determined with the viscosity of water, which is much lower than the oxygenated fluid viscosity. Therefore, the resistances for each section were calculated in order to proceed with the analysis.

3.1.2 Resistances

By calculating the resistances from each section of tubing it was possible to refine the model above to more accurately reflect the pressure drops and eventually how much force is needed to

achieve the desired flow rate. Resistances were calculated by simplifying the Hagen-Poiseuille equation. If

$$\Delta P = RQ \tag{3.2}$$

then the resistance is,

$$R = \frac{8\mu L}{\pi r^4} \tag{3.3}$$

Before calculating the resistances, it was important to select the correct viscosities for each section. In section A, the viscosity of the fluid saline was used (which was approximated to be the about the same as water) but for sections B and C, it was necessary to find the viscosity of the fluid at a concentration of both 90% and 70%. The viscosity of the fluid at different concentrations was found by using a rheometer and the data provided by doctor John Kheir is shown in Figure (3.2).

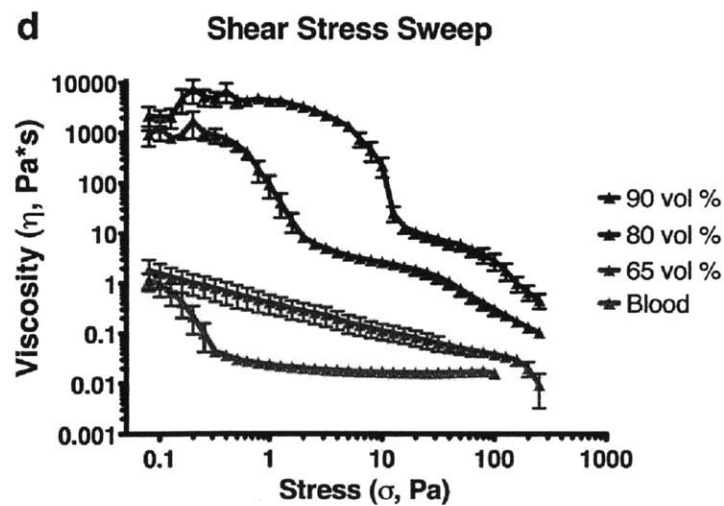


Figure (3.2) – Viscosity of oxygenated fluid at different concentrations found using a rheometer.

Using the chart in Figure (3.2), the viscosity of the oxygenated fluid at a 90% concentration was approximated to be around 10 Pa*s and at 70% concentration it was approximated to be around 0.1 Pa*s. Imputing these new viscosities into the model and using equation (3.3), the following resistances were calculated shown in Table (3.2)

Table (3.2) – Resistances calculated for each section using equation (3.3) and the viscosities estimated from figure (3.2).

Section	A	B	C
Resistance (Pa*s/m ³)	17,822,021	89,110,103,498	29,712,636,326

With the resistances calculated above, it is possible to calculate how much force will be needed to achieve the flow rate necessary if the area of the top of the bellows is known. The bellows selected hold slightly more than 1L of fluid. It has a diameter at the top of 115mm, a height of 150mm, and an inner cylinder of diameter 90mm (for more information on bellows selection and calculation refer to section 3.2.4). Using the equation,

$$F = P \times A \quad (3.4)$$

The force needed to compress the bellows and achieve the flow rate desired was about 700 lbs or around 3300 N. A motor that could achieve this force and still have the resolution and control needed for this application was not readily available. Therefore, the model was changed in order to decrease the resistances in each section. By examining Poiseuille's equation, it was determined that the best way to decrease the resistance would be to increase the radius of the tubes in each section. Since, the viscosities and flow rates are fixed, the only other parameters that could be changed are length and radii of each section. However, decreasing the length of each section is not as affective as changing the radii because the radius is a quadratic term. By

changing the radii and decreasing the lengths of different tubing sections it was possible to decrease the force needed to achieve the desired flow rate to around 60 lbs or about 250 Newtons. The new parameters and calculations are shown in Table (3.3).

Table (3.3) – Revised analysis performed using Poiseuille’s equation.

	Section		
	A	B	C
Length (m)	0.6604	0.1397	0.8669
Radius (m)	0.003175	0.0047625	0.003175
Q (m ³ /sec)	8.33333E-07	2.5E-06	3.33333E-06
μ (Pa*sec)	0.001	10	0.1
P_final (Pa)	7,241	7,241	-
ΔP	13.79	17,288	7,241
P_initial (Pa)	7255.04	24,529	7,241
ΔP (psi)	1.05	3.56	1.05
Force (N)	N/A	255	N/A
Force (lbs)	N/A	57	N/A
Resistance	16,549,019	6,915,069,760	2,172,372,011

By comparing tables (3.1) and (3.3) it is shown that changing the radii and lengths of tubing significantly reduced the resistance in sections B and C. With a force of 60 pounds, the design process can now proceed and the appropriate motor and load cell can be selected.

3.2 PUMP DESIGN

The pump design described in this section was carefully constructed to meet all of the functional requirements mentioned in Chapter 2. The basic structure of the pump was designed based on the fluid modeling in section 3.1. This section will use that fluid model to describe some of the decisions made in selecting the components. It will also go into detail and analysis of different components of the pump design and the reasons behind the choices made.

3.2.1 Structure

The frame and structure of the pump was built using 80-20 structural elements. The 80-20 was selected because it would allow for a very modular design that can be easily assembled. Since the bellows had a diameter of about 120mm or about 4.7 inches, the frame was designed to be 8 inches wide so that there was about 2 inches on either side of the bellows. It was important to have a reasonable gap on either side of the bellows to facilitate loading. The top of the structure has an 8 by 8 plate $\frac{1}{4}$ " thick in which the electronics and motor are mounted to. The motor is mounted in such a way that the lead screw is attached to a plate that is guided by the rails on the 80-20 and push down on top of the bellows. The bellows sit on a plate, which is fixed to the 80-20 rails slightly above the center of the structure. There is also another structure off to the side of the main structure in which the peristaltic pump sits on. The basic structure of the pump is shown in figure (3.3).

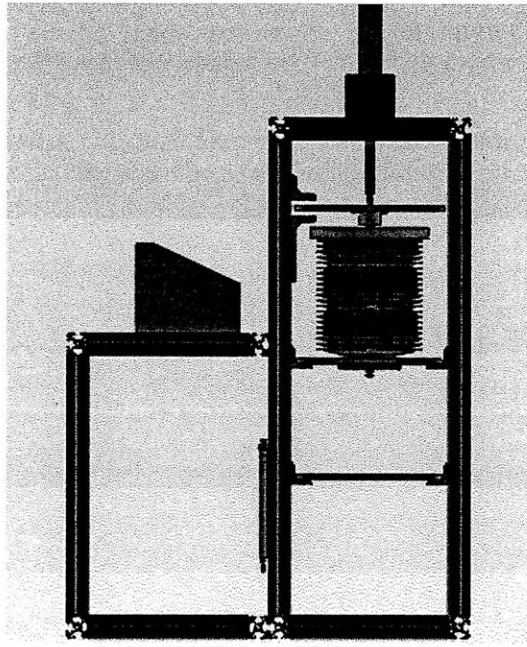


Figure (3.3) – Basic structure of the oxygenated fluid pump

3.2.2 Tubing and Connections

Most of the tube sizing was determined in the fluid model in section 3.1. The lengths used in the model were adjusted so that all of the components were connected. The tubes were connected using different push to connect junctions. The tube connected to the bellows pump meets the tube connected to the peristaltic pump at a wye junction. The tubes then go on to connect to the static helical mixer and to the bubble trap. All the components are shown in Figure (3.4) and listed on table (3.4).

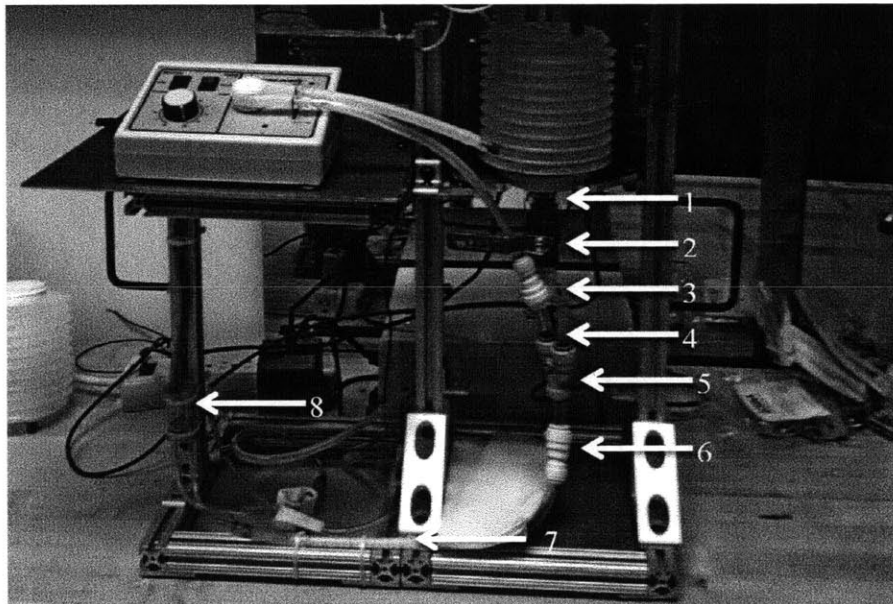


Figure (3.4) – All the plumbing components and connectors

The plumbing components are labeled in Figure (3.4) above with numbers 1-8 and are listed in Table (3.4) below:

Table (3.4) – Outlines all the plumbing components and their dimensions shown in Figure(3.4).

Labeled Number	Component	Important dimensions
1	Straight thread to Pipe Thread	$\frac{3}{4}$ "-16 UNF-2A to $\frac{1}{2}$ " NPTF
2	Ball Valve	$\frac{1}{2}$ " NPTF to $\frac{1}{2}$ " MIP
3	Pipe Thread to Quick Connect	$\frac{1}{2}$ " MIP to $\frac{1}{2}$ " Tube
4	Tubing	$\frac{1}{2}$ " Tube
5	Wye Quick Connect	$\frac{1}{2}$ " Tube
6	Quick Connect Reducer	$\frac{1}{2}$ " Tube to $\frac{3}{8}$ " Tube
7	Helical Mixer	N/A
8	Bubble Trap	N/A

3.2.3 Motor & Load Cell Selection

Both the motor and the load cell were selected based on the force calculation determined from the fluid model. The force needed to achieve the desired flow rate of 200mL/min as specified by doctor John Kheir is about 60lbs. The motor selected for this application was a stepper motor that can operate at a maximum force of 100 lbs and has a step size of 0.001". The force vs. linear velocity curve for the motor can be found on appendix (). The load cell selected was an Omega LCFD-100 that can sense forces up to 100 lbs also. It is attached to an amplifier, which outputs a signal large enough to be read by the LabView DAQ board.

3.2.4 Bellows

Bellows are a good choice for this application because of the different functional requirements it had to fulfill. It can hold slightly over 1L of fluid and has a predictable force profile that can be

easily measured and modeled as a spring. Commonly, syringes are used for medical pumps however syringes are rarely found with a 1L capacity and the friction force generated by the seal in the syringe is complex to model. The bellows is also a good choice because of the volume compression method that is being used. As mentioned before, the bellows has a diameter of about 120 mm, a height of about 140 mm, and an inner cylinder of diameter 100 mm.

Some slight modifications had to be added to the bellows. On the top of the bellows, a small luer lock is attached and acts as a purge valve. Once the bellows is full of fluid, the top can open so that the excess air can be purged out of the bellows. Also, the bottom of the bellows is open and threaded, therefore, a 3-D printed cap was made in order to close the bellows and create a container. The cap was specifically designed to create a seal between the bellows and the cap. Adding a cap also made it easier to load up the bellows with the oxygenated fluid. Figure (3.5) below shows the bellows with the cap and purge valve.

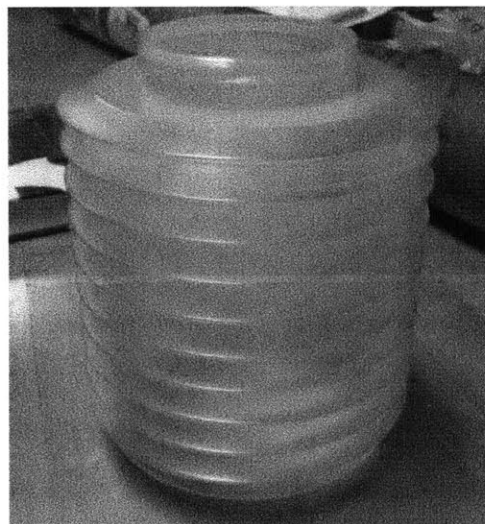


Figure (3.5) – Bellows with purge valve and 3-D printed cap.

3.2.5 Bellows Cap

A cap was designed for the bellows in order to make it a closed container. It was designed so that it would thread into the bottom of the bellows and create a perfect seal. To create the proper seals, o-rings were incorporated into the design. Figure (3.6) shows the cap design,

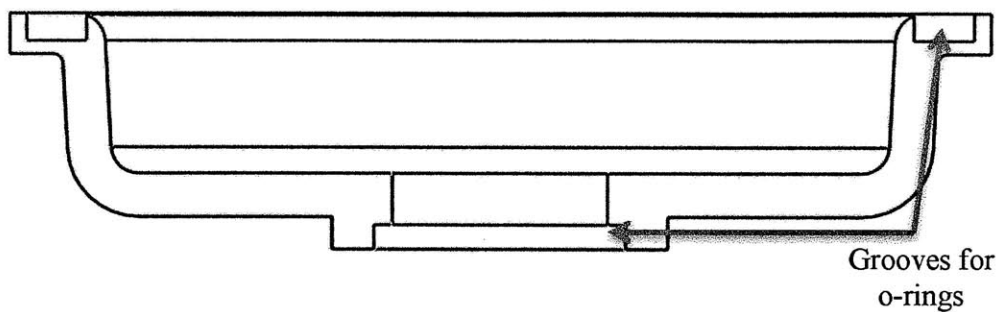


Figure (3.6) – This figure shows the cap design with grooves for the O-rings.

In designing the grooves for the o-rings it was important to dimension the part correctly so that the ring has space to deform. Figure (3.7) more closely shows what is happening.

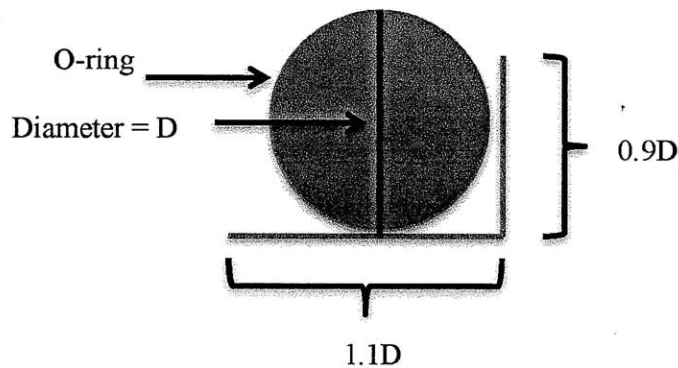


Figure (3.7) – This is the geometry necessary to create a seal.

In figure (3.7), it is show that the height of the groove has to be 90% of the o-ring diameter and the length has to be 110% the diameter of the o-ring. This is done so that when the o-ring is

compressed, it will deform and push against the side of the groove, creating a perfect seal. This technique was used in two places to create a seal between the bellows and the cap, and to create a seal between the cap and the straight-thread to NPT connector.

The final design of the bellows cap is shown below in Figure (3.8). The image on the left is the CAD model from SolidWorks designed by Alex Mason to thread into the bottom of the bellows. The image on the right is the actual bellows cap after it was 3-D printed. The initial cap was created using a 3D printer provided by MIT's Center for Bits and Atoms, and 3D Printsmith LLC printed a second version. Large-scale production would use injection molding.

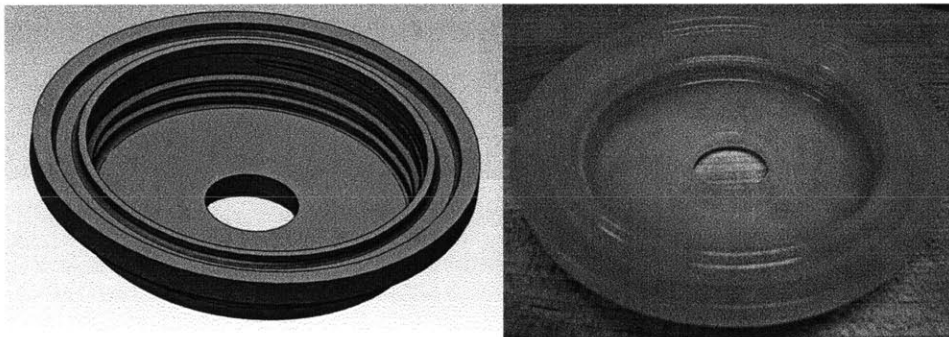


Figure (3.8) – 3D CAD model of the cap on the left along with the actual 3D printed part on the right.

3.2.6 Volume Calculation

Calculating the exact volume of the bellows was a really important step in calculating how that volume changes as it is being compressed. The total volume of the bellows was determined to be 1.28 liters. This volume is more than the specified 1L because when the bellows is fully compressed, there is still some fluid inside it. This is because the bellows can compress a maximum of about 83mm. This means that when the bellows is fully compressed, it is still 57mm high and holds a volume of about 275mL. This means that the bellows dispenses a total

volume of about 1005mL, which is right around what is needed to dispense due to our functional requirement. In reality, the bellows needs to displace only around 750mL of fluid because the functional requirement states that 1L of the fluid needs to be dispensed but that is at the 70% concentration. Since the bellows is holding the 90% concentrated fluid, it still has to be mixed with saline, which is where the other 250mL comes in.

After measuring how much fluid the bellows can hold, the volume displaced as the bellows is being compressed was calculated. This calculation took several steps. The first step was calculating the volume displaced as the inner cylinder of the bellows got smaller. Figure (3.9) demonstrated this section.

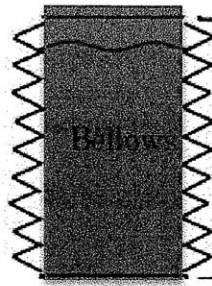


Figure (3.9) – The shaded blue portion indicates the inner cylinder of the bellows.

Calculating the volume of the shaded portion in Figure (11) was done by using Equation (3.5) below:

$$V_{cylinder} = \pi r^2 L \quad (3.5)$$

where r is the radius and L is the displacement of the bellows. The radius of the center cylinder section was measured to be 50mm. The length of the bellows was measured to be about 130mm giving the center section a volume of approximately 1.02 liters. To model the volume displaced as the bellows is being compressed, L was turned into a variable. The next step was to calculate

the volume of the un-shaded portion of Figure (3.9). This part was much more complicated. The ridges were modeled as triangles as shown in Figure (3.10):

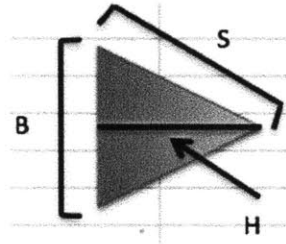


Figure (3.10) – Triangle used to model the cross section of the ridges of the bellows.

In Figure (3.10), B is the base of the triangle, H is the height and S is the side. As the bellows is being compressed, the base of the triangle changes. Since the side (S) is a constant dimension of the bellow, then the height of the triangle also changes as the bellows is being compressed. To simplify calculations, only half of the triangle was used and then was multiplied by two. Using this cross sectional area, the shell method show in equation (3.6) was used to calculate the volume of the ridges.

$$V_{ridges} = 2\pi \oint r(f(x))dx \quad (3.6)$$

$f(x)$ is the equation determined by the line with a slope of S. The limits of integration were determined by the radius of the bellows from the center until the outside of the bellows or as the radius goes from r to $r + H$. $f(x)$ was determined by equation (3.7) below:

$$f(x) = -\left(\frac{B}{2H}\right)x + Y \quad (3.7)$$

where Y is the y-intercept. For example, Figure (3.11) bellow shows an example of the graph created when B is 5.9mm, H is 12.5mm and Y is 28.32mm. Y was determined by using the linear

equation $y = mx + b$ and the x-intercept of 60 because that is the maximum radius and that is achieved when $B/2$ is zero.

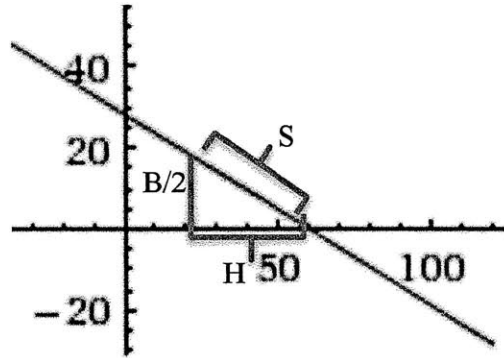


Figure (3.11) - An example of the cross-sectional area that was used to find the volume of the ridges on the bellows.

H was found by using the equation of a circle with S being the radius and the center of the circle set at $x = 60$ and $y = 0$. This is shown in Figure (3.12) below:

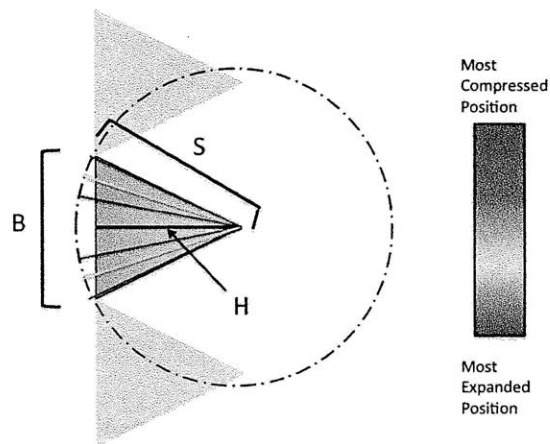


Figure (3.12) – Demonstrates how H changes as B is getting smaller, representing the bellows being compressed.

By repeating this method as B decreases to zero and as H approaches S, then multiplying to account for all the ridges, the volume displaced by the ridges turned out to be around 240mL.

Adding this to the volume displaced by the center cylinder the total volume displaced was calculated to be about 1.26L, which is extremely close to the measured volume of 1.28L. This approximation can get closer by measuring the dimensions of the bellows with more precision however because the flow rate needs to be accurate to $\pm 10\%$, this approximation is sufficient.

3.2.7 *Sliding Mechanism*

The sliding mechanism is the most important mechanical component of the pump. It is what aligns the lead screw of the motor and allows the bellows to be compressed in a vertical motion with the force being straight down on top of the bellows. The sliding mechanism consists of a few components shown in Figure (3.13).

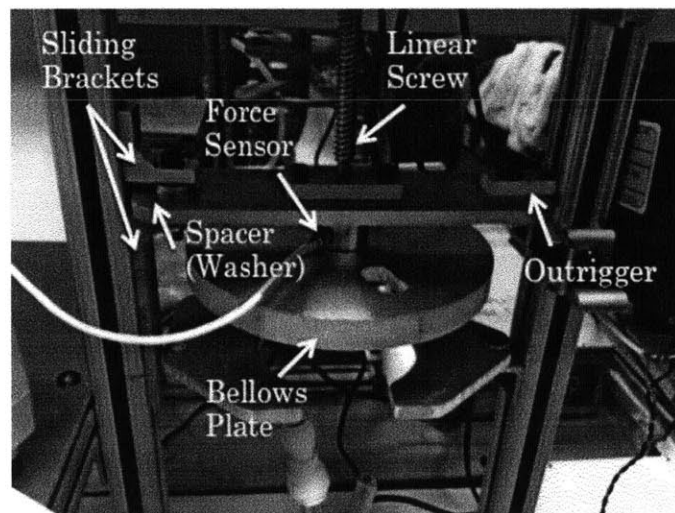


Figure (3.13) – Sliding mechanism and its corresponding components.

The geometry of the sliding mechanism was the most important part of its design. The distance from the top of the upper sliding bracket to the bottom of the lower sliding bracket has to be at least 1.6 times the distance from the left rail to the linear screw. This needs to be true because it has to obey St. Venant's principle where L/D is at least 1.6 to avoid jamming where D is the

characteristic length and L in this case is the length of the sliding brackets. D was measured to be around 94 mm and L was determined to be about 150 mm in order to fulfill St. Venant's principle. If up and down makes up the y-axis, left and right make up the x-axis and in and out of the page the z-axis, then the sliding brackets prevent motion in the positive and negative x and z directions as well as rotation around the z and x axes or 4 degrees of freedom. The outrigger on the other side prevents rotation around the y- axis. The only degree of freedom left is travel in the y direction, which is the intended direction of travel. This mechanism relies on the guiding rails on either side to be as parallel as possible or else jamming would occur. The sliding mechanism designed here works well and was an easy, reliable way to compress the bellows.

3.2.8 Peristaltic Pump

Saline is pumped by a Fisher Scientific Variable-flow peristaltic pump, which is controlled manually. The peristaltic pump is capable of producing flow rates from 30mL/min to 100mL/min, which covers the required range. The pump is a simple device that is familiar to doctors and nurses. We decided to use a peristaltic pump because of its ease of use and its minimal contact with the fluid. One problem encountered however was that the peristaltic pump used was not reliable. With a saline bag standing upright, the flow would vary greatly on the same setting because of the difference in pressure as the level of saline drops. Therefore, the saline bag was laid on its side and the pump functioned more reliably. Ideally, a better peristaltic pump would be purchased.

3.2.9 Static Helical Mixer

The final module of the design is the mixer, which combines the oxygenated fluid with saline. This reduces the concentration from 90% to 70% gas by volume, which in turn reduces the viscosity of the fluid and makes it suitable for infusion. The concentrated oxygenated fluid and saline meet at a wye-junction to minimize damage to the fluid. They then combine and pass through a single plumbing path into a static, helical mixer, similar to an epoxy mixer.

CHAPTER

4

PUMP LOGIC

4.1 LOGIC

The novelty of the oxygenated fluid pump is in the control system. It is broken down into three main phases: a purge phase, a calibration phase and a motor control phase. The logic of the pump is displayed in the diagrams below:

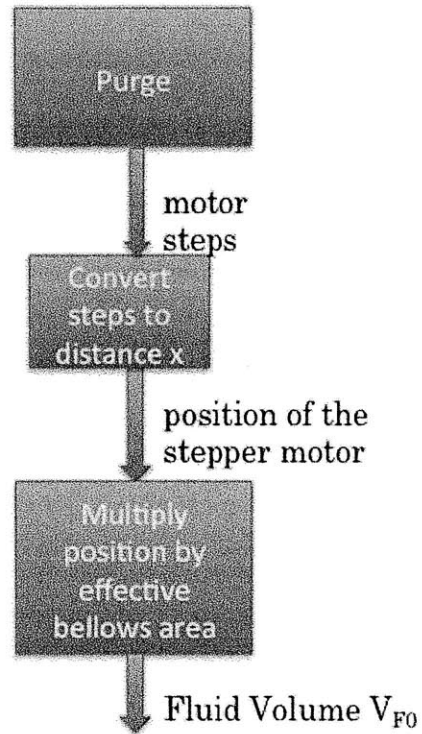


Figure (4.1) – Pump logic diagram: Purge Phase

The purge phase is utilized to purge the bellows of all unwanted air. It is assumed that while loading the bellows with fluid, some air will be trapped. This first step will get rid of the air and calculate a new volume which is set as the initial volume of fluid in the bellows. The next step is calibration, which is displayed below:

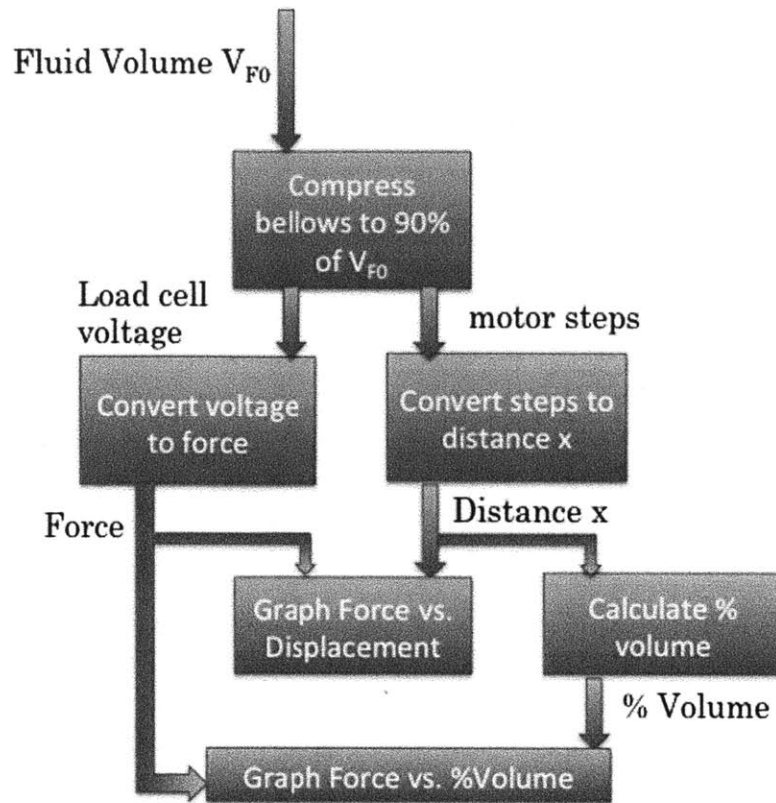


Figure (4.2) – Pump logic diagram: Calibration Phase

The calibration phase is necessary because the fluid properties are not consistent every time. Also, the bellows may not be filled the same every time or some of the fluid may have degraded faster than others. The calibration will first ensure that the fluid properties are within reasonable range then determine the compressibility constant for that particular batch before starting to pump. Right now, the calibration phase is being used to help characterize the fluid. The motor control phase is completely dependent on this calibration. Ideally, once the fluid is characterized there will be a model used for an ideal fluid in which case the calibration won't be necessary anymore. If anything it will be used to categorize the fluid as ideal or non-ideal. If the fluid is non-ideal then another batch will have to be used.

The calibration phase works by taking the initial fluid volume and calculating 90% of that volume. Then the number of steps the motor needs to travel is calculated. The motor goes to this set position and records the force on the load cell at every step and makes a graph. This graph is basically the spring force the fluid exerts as it is being compressed.

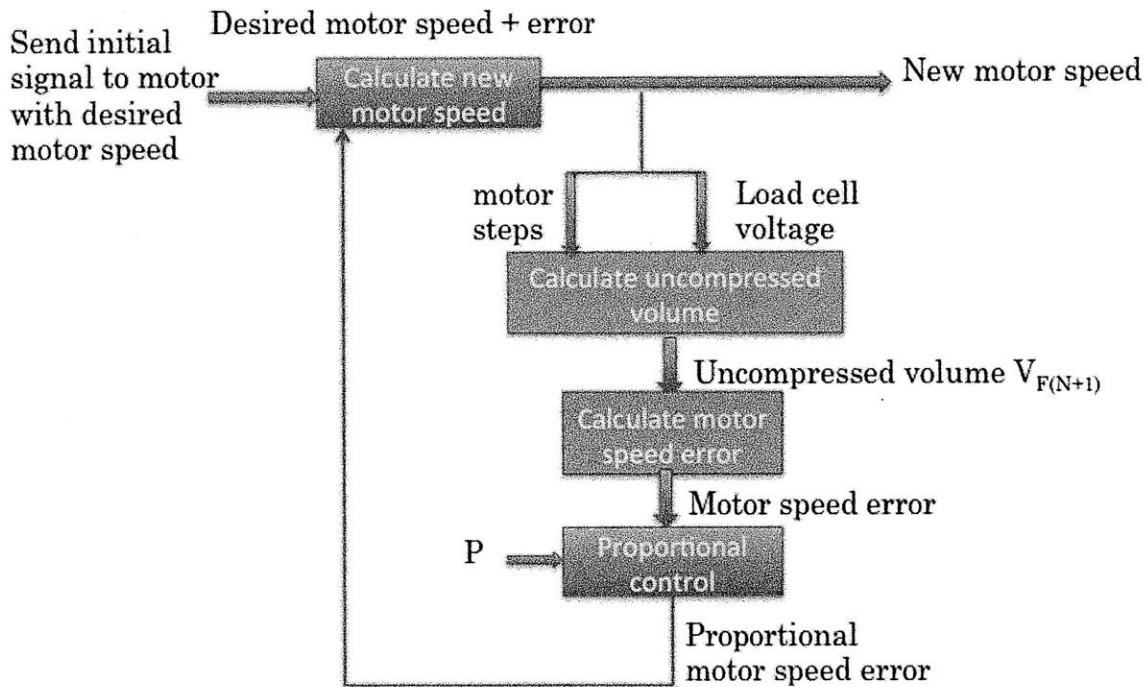


Figure (4.3) – Pump logic diagram: Motor Control Phase

The motor control phase is the most important phase in the control system and the most complicated. As the motor is moving down, the force is being measured. This force is then used and compared to a force that matches it closest in the calibration graph. Then, a corresponding volume is calculated using this reading, which is then used to calculate a flow rate. This flow rate is compared to the desired flow rate (set by the user) and an error is calculated. This calculated error is then converted into an error in velocity. The error in velocity is finally added to the

current velocity of the motor, which will speed up or slow down depending if the error is positive or negative. The load cell then reads another force and the process is repeated. This phase is better outlined in the Labview section 4.2.5, where the controls are explained in detail.

4.2 LABVIEW

The code was written in Labview using a phidgets controller and a DAQ board. A phidgets library was downloaded from the phidgets website. The software progresses through a number of stages; in order, they are “Initialize”, “Bellows Top”, “Manual Jog”, “Calibration”, and “Motor Control”, and “End”.

- “Initialize” checks connections and makes sure the stepper motor is ready to be operated.
- In “Bellows Top”, the stepper motor moves downward until a non-zero force is detected. This indicates the top of the bellows.
- “Manual Jog” allows the user to manually control the position so that all the air from the bellows is purged before calibration.
- The “Calibration” stage performs a test compression of the fluid and determines its compressibility curve, as this may vary from batch to batch.
- The “Motor Control” stage dispenses fluid with a force feedback system.
- “End” cleans up the program and closes the operation of the stepper motor.

The next few sections will outline each of the six steps outlined above in more detail. The six stages above are inside a case structure. A case structure has one or more sub-diagrams that execute once the structure executes. This case structure is inside a while loop, which repeats the sub-diagram inside it until another value indicates it to stop. In this case, a stop button will stop

the loop. Outside the while loop are channels that only need to be open once. For example, it opens up the channel to communicate with the motor through a phidgets controller. It also opens up the limit switches and the load cell reading. Once these channels are open, they can be utilized at any point in the rest of the code. The only other component outside of the while loop is the close function, which closes communication to the components that were opened. The rest of this chapter will go on to explain each of the six cases inside the case structure.

4.2.1 Initialize

The purpose of this stage is to raise the bellows plate to its highest position to allow for easy loading of the bellows. The first block sets the velocity limit of the motor to any set constant value. The second block sets the target position. The way this works is that a shift registry is used. The first thing that needs to be known is that this case is going to be repeated inside the while loop until it is told to move on to the next case. The shift registry remembers the last set input of a block and carries it back around the while loop. Using this knowledge, one step is added to the motor position every time the loop runs. For example, when the case executes for the first time, the default target position is one, then the shift registry remembers that value. The second time the case is executed, the shift registry remembers the value of one and adds one to it to make it a value of two. This value is then set as the new target position. This way, the motor will continue to increase step by step until it is told to stop.

There is only one caveat; the code should not run faster than the speed of the motor. The velocity limit has to be the limiting factor in how fast the motor moves. The reason for this is that if the code is running faster than the motor speed, when the command to stop is executed, the motor will not be at the set target position and will continue to move until that position is reached. If the

motor velocity is the limiting factor though, as soon as the stop command is executed, the motor will stop moving.

A limit switch in this case activates the stop command. As long as the limit switch reads a negative value, the motor will continue moving the bellows plate up. Once the limit switch is pressed, the motor will reverse its direction and move until the limit switch is un-pressed. This is done by using a true/false case structure. A comparison block is used to compare the value of the limit switch to a “true” Boolean value. Once the value of the limit switch is true, or when it is pressed, the code will execute the sub-diagram inside the true/false case structure. When the case is false, nothing happens. When it is true, a new target position is set which will move the motor plate down until it the limit switch is no longer active. The true case structure also tells the code to move to the next case, which is the Bellows Top case. Last but not least, there is a wait function inside the true/false case structure that waits until the motor gets to the un-pressed limit switch position.

4.2.2 Bellows Top

The purpose of this stage is to find the top of the bellows. Much in the same way the initialize state works, a value of one is added to the target position using a shift registry until it is told to stop. The only difference is that the motor is moving the bellows plate down instead of up. The stop command in this case is also different. Instead of a limit switch activating the comparison block it is the limit switch value. The comparison block is once again connected to a true/false case structure. If the load cell reads a value greater than a specified value greater than zero, it executes the sub-diagram inside the true case of the true/false case structure. Once again, the false case is just empty, allowing the motor to continue moving down. When the comparison

block reads a true value however, the true case is activated. The true case simply tells the motor to stop and moves the code to the Manual Jog case. Therefore, the Bellows Top case moves the motor until the load cell reads a value greater than zero, in which case it stops and the top of the bellows is found.

4.2.3 Manual Jog

The goal of the Manual Jog case is to allow the user to manually control the position of bellows top. In this state, the purge valve will be opened, then the user can manually slide a control and compress the bellows until all of the air is purged from the bellows. At which case the purge valve is closed and calibration can occur. The first step in this case is to set the current position to zero. This needs to be done because once the bellows is purged, this position sets the initial fluid volume that will be used in other calculations down the road, including the error calculation. This position is set to zero by using yet another true/false case structure. This case structure is set to a default false value. A true/false switch, also set to a default false value, activates it. Once the case is executed once, the position is set to zero and the value on the true/false switch changes to true. The true case of the case structure is just empty, allowing the code to continue.

The next step in the code is an event structure. An event structure has one or more sub-diagrams or event cases, one of which executes once the structure executes. The default event just gets the motor position. It does not actually move the motor, it just tells the user where it is from the zero position. The second event in the event structure is just a block with the target position, which is controlled by a slider. When the slider moves, it tells the motor to go to that position. When the manual jog is complete, a button tells it to go to the next case, which is the Calibration case.

When the button is not pressed, the Manual Jog state just keeps running. Once the button is pressed, the calibration state is initiated.

4.2.4 Calibration

The calibration state is one of the most important states. The goal of the calibration state is to compress the bellows and get a calibration curve or a compressibility curve for the fluid. It does this by first finding the position of the bellows. Then it takes the manual jog position, calculates how many steps need to be taken in order to compress the bellows to 90% of its volume after the manual purge. It does this math using a sub-VI or sub virtual interface described in detail in Appendix (C).

The number calculated is then entered as the number of iteration for a for loop. A for loop executes its sub-diagram n times where n is value entered by the user, in this case, the number of steps needed to compress the bellows to 90% of its volume. 90% was chosen because compressing the bellow to 90% of its original volume generates a force large enough to set a limit on the force. During execution of the motor control, the force should never go above the force needed to compress the bellows to 90% of its volume.

Once the number of iterations is set, a shift registry takes the iteration that it is in, then adds that number to its position after the manual jog. Therefore, if the motor is in position 50 after the manual jog and the for loop is in its first iteration, then the motor is set to a target position of 51. During the second iteration the target position is set to 52 and so on. These positions are also being recorded into an array, along with a force reading at every step. Last but not least, inside

the for loop, the program is set to record force values only if they are larger than its previous value, making it monotonic.

Once the for loop is complete, the values are inserted into a sub-vi that calculates a %volume. This percent volume is basically what volume the compressible fluid will expand to at atmospheric pressure. This calculation is also explained in Appendix (). After the %volume is calculated, a graph of %Volume vs. Force, which is referred back to in the motor control stage is created. Last but not least, the motor is set back to its starting position, after the manual purge. Another button tells the code to proceed into the motor control state.

4.2.5 Motor Control

The motor control phase use a force feedback system in order to control the motor. The first step it does is set a target position to the bottom of the stroke of the bellows. A position that will guarantee the bellows becomes fully compressed. The motor control state functions around a sub-vi, which calculates the motor speed error. This sub-vi has six inputs.

One of the inputs is found when the load cell reads a force. This force is rounded to the nearest integer. The force vs. % volume graph from the calibration state is also rounded to the nearest integer. The code then compares the force read to the array from the calibration state and finds a value that matches the force exactly. Once this force is found, the index is recorded and the corresponding %volume is displayed. This value is then inserted into the sub-vi that calculates the error. If the force is not matched exactly, a shift registry makes sure that the previously matched force is inserted instead.

Another input is the amount of steps a motor moves in one loop. This is found by getting the current position at the beginning of every loop, and using a shift registry to subtract the previous position recorded. This change in position is also used to calculate another input. The current velocity of the motor is read and the number of steps moved in each loop is divided by this current velocity in order to calculate a time step for each loop. The last three inputs are the desired flow rate, the uncompressed volume calculated at a previous step (initial value is 0, later values are determined by using a shift registry) and the manual jog position.

These inputs are used in the following order to calculate an error in motor speed. The manual purge is used along with the number of steps traveled in each loop are used to calculate a volume of fluid displaced. That number is then divided by the %Volume input in order to find the volume at atmospheric pressure. This volume is then subtracted by the previously calculated uncompressed volume (in the first iteration it is zero). This volume is then divided by the time input in order to find a flow rate. This flow rate is then compared to the desired flow rate input and an error is calculated. This error is then divided by an effective area of the bellows in order to get an error in velocity.

A small constant multiplies this error in order to create a proportional control. If the error is negative, it will slow the motor down, if it is positive then it will speed the motor up. An extra measure of safety is also added. The previous velocity measured is multiplied by 120%. If the new velocity calculated after the error was added to it is greater than 120% of the old velocity, a new velocity is not set. This will ensure that the pump does not increase its speed too quickly.

The motor control is only stopped when a limit switch is pressed. It works in the same manner as the previous limit switch in the initiate case, by using a true/false case structure. Once the true/false case structure achieves a true value, the motor position is set back up to un-compress the bellows and the code moves on to the final end state.

4.2.6 End

The end state is just a blank sub-diagram that waits for the motor to move back up until the bellows is uncompressed. Once the motor reaches its target position, the code will just continue running a blank diagram until the stop button is pressed, in which case communication to the motor, limit switches and load cell are closed and the code stops running.

EXPERIMENTATION

5.1 FLOW RATE

The flow rate for both the peristaltic pump and the bellows pump needed to be determined experimentally. It is essential that the pump functions properly and delivers the flow rate with a $\pm 10\%$ maximum error. Some experiments are described below, showing the reliability of the flow rates of each of the pumps.

5.1.1 Peristaltic Pump

For the first test, the flow rate of the peristaltic pump will be measured. This test will ensure that the pump is reliable and will pump out a consistent amount of saline every time. Three minutes was chosen as the test time because the pump will be pumping for a total of 5 minutes at a time (5 minutes is the time we will deliver 1L of mixed fluid) and figured that if it pumps steady for 3 minutes then it will be fine with 5. The main concern is that the pump needs to be consistent in order to deliver well-mixed fluid. The flow rate has to be the same –whether the bag is full or half full.

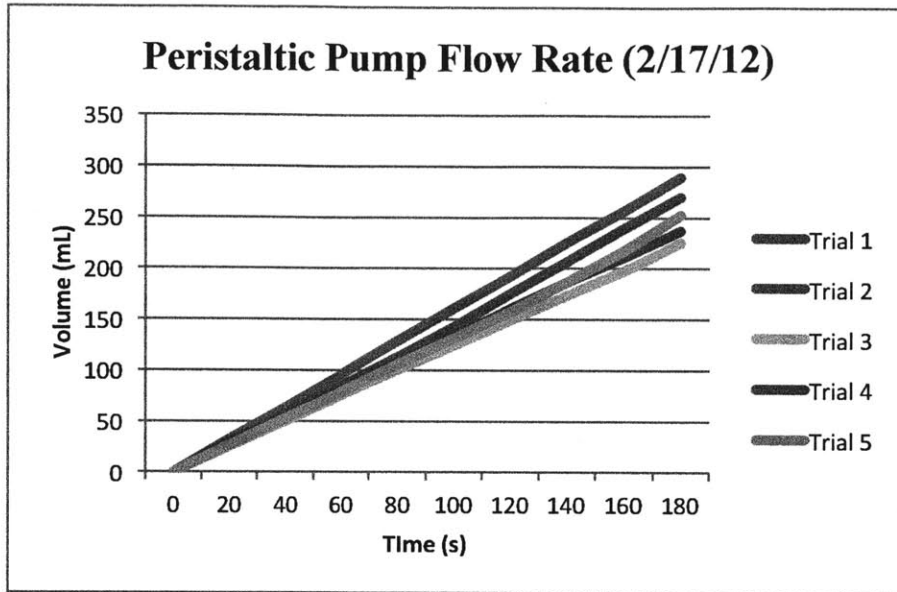


Figure (5.1) – This figure shows the flow rates measured for the peristaltic pump on 5 different trials

These test resulted in an average flow rate of 84.9 ± 10 mL/min, which is not accurate enough for our system. The flow is linearly increasing; however, there is too much deviation between each trial considering the input setting/speed of the pump was the same. It turns out that as the volume in the saline bag changed, the flow rate changed. As stated in the functional requirements, the flow rate needs to be accurate to $\pm 10\%$ and has to be consistent whether the bag is full or not. A different experimental setup was then attempted in which the saline bag was horizontal with an aluminum plate on top of it. An average flow rate of 85.5 mL/min was found with a standard deviation of 2.4 mL/min. This test ensured that the pump is reliable and will pump out a consistent amount of saline in every use if the saline bag is kept horizontal.

5.1.2 Bellows Pump

The flow rate of the bellows pump will be tested by measuring the volume of fluid dispensed over 20 second time intervals for a total of 3 minutes. This test will reveal how reliable the pump

is in terms of consistently delivering a specific volume of fluid. All these initial tests will be done with water. The input setting on the pump will be constant for this round of data collection.

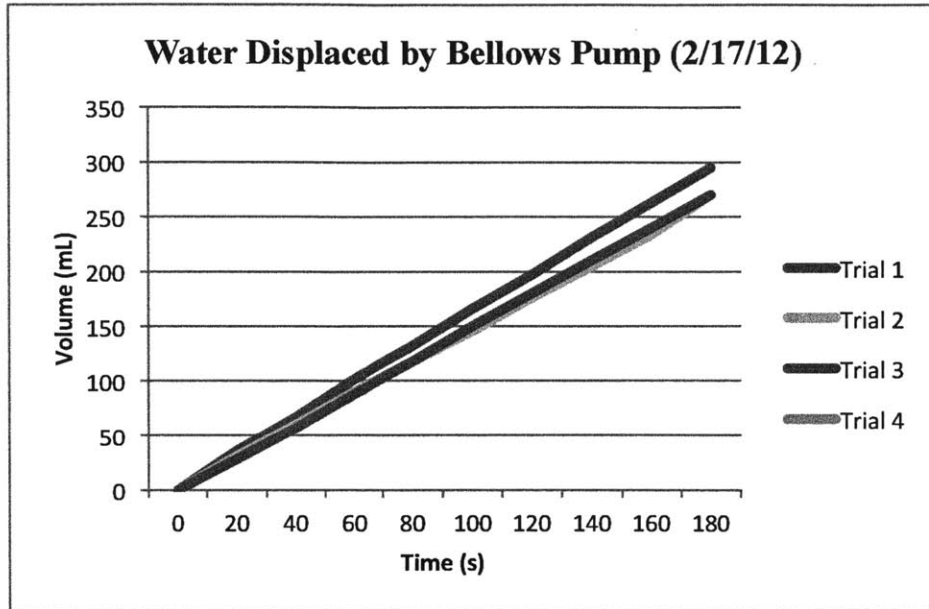


Figure (5.2) – This figure shows the flow rates measured for the bellows pump on 4 different trials
 These test resulted in an average flow rate of about 96mL/min. The flow rate for the bellows pump is much steadier than the peristaltic pump. Taking a closer look at the flow rates for each trial in Figure (5.3) below:

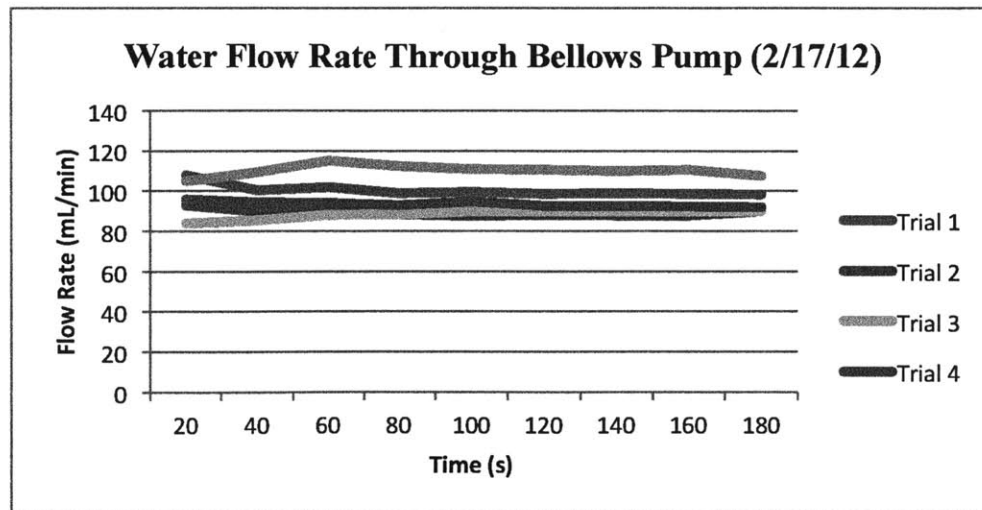


Figure (5.3) – This figure shows the flow rates at different times during pumping

Once again, plotting the flow rate at different points gives more detail on how the pump is behaving. The flow rate varies by no more than about 15mL/min, which is promising. The bellows pump is much more reliable than the peristaltic pump. Further testing needs to be done with both the peristaltic pump and the bellows pump to determine if the pumps will behave the same with a viscous, compressible fluid.

5.2 PRIME TIME

Prime time is how long it takes for the flow to go throughout the system before it actually starts dispensing fluid. The prime time was measured for the peristaltic pump but nothing conclusive was found because the peristaltic pump was unreliable. An important step before moving along would be to repeat these tests, making sure to record the prime time also.

FLUID QUALITY

6.1 QUALITY OF FLUID¹

Evaluating the quality of the delivered fluid is paramount for this device. Concerns include: volume percentage, particle size distribution, oxygen tension, rheometry and degradation. Since the 90 percent gas by volume, oxygenated fluid is compressible and non-newtonian, it may be helpful to assess the fluid at a range of concentrations—from 10 percent to 90 percent gas by volume in 20 percent intervals. This would allow for easy identification of at what concentration any problems arise.

6.2 VOLUME PERCENTAGE

It is important to confirm that the mixed fluid dispensed by the pump is 70% gas by volume, as it is one of the functional requirements requested by the client. In order to do this, a defined volume of fluid within a syringe can be weighed to calculate the volume of the fluid phase, assuming that the fluid is mostly water so its density is known ($\rho_{water} = 1g / mL$); the mass of an empty syringe with a cap is 74.56 grams and must be accounted for in calculating the mass of fluid alone. The remainder of the volume contained within the syringe can then be assumed to be gas trapped

¹ All information in Chapter 6 was listed with extensive help from Kristen Peña.

within lipidic oxygen microparticles (LOMs) and the volume percentage can be calculated using Equation (6.1),

$$Volume\% = \frac{V_{gas}}{V_{total}}$$

$$Volume\% = \frac{V_{total} - V_{liquid}}{V_{total}} \quad (6.1)$$

$$Volume\% = \frac{V_{total} - [(m_{total} - m_{syringe}) / \rho_{water}]}{V_{total}}$$

The oxygenated fluid originally provided by the doctor is always about 90 percent gas by volume but since Dr. Kheir's fluid manufacturing procedure is not entirely consistent, the volume percentage must be verified first using the method just detailed. This original volume percentage can then be used to determine the amounts of oxygenated fluid and saline needed for a certain desired volume percentage. Equation (6.2) allows for calculation of the volume of oxygenated fluid need for a certain desired volume percentage and total desired volume, while Equation (6.3) allows for calculation of the volume of saline needed in the mixture.

$$Volume\ O_2\ Fluid(mL) = \frac{Desired\ Volume\ \%}{Original\ Volume\ \%} \cdot Total\ Desired\ Volume\ (mL) \quad (6.2)$$

$$Volume\ Saline\ (mL) = Total\ Desired\ Volume(mL) - Volume\ O_2(mL) \quad (6.3)$$

The results of using Equations 6.2 and 6.3 to calculate the volumes of oxygenated fluid and saline needed for the range of concentrations detailed are displayed in Table 6.1. In this case, the original oxygenated fluid provided by the doctor was determined to be 93 percent using the weighing method.

Table (6.1) - Theoretical Mixing Proportions

Desired Volume %	Volume % Original	Volume O2 (mL)	Volume Saline (mL)
10	93	10.75	89.25
30	93	32.26	67.74
50	93	53.76	46.24
70	93	75.27	24.73
90	93	96.77	3.23

Mixtures were made by first drawing the desired volume of oxygenated fluid from the original syringe into a new one using a T-connection with luer locks. Similarly, the desired volume of saline was drawn into the new syringe, and the two fluids were mixed by hand agitation. The mass was determined using the Mettler Toledo scale shown in Figure (6.1) and the volume percentage was confirmed using Equation (6.1).



Figure (6.1) - Mettler Toledo Scale

Unfortunately, when this theory was tested in lab, these predictions resulted in over-dilution of the oxygenated fluid, so a smaller volume percentage than intended. For example, mixing 11 mL of the 93 percent oxygenated fluid with 89 mL of saline for the intended 10 percent gas by volume mixture, resulted in a mass of 85.5 g and volume percentage of 4.96.

Since the theoretical model developed did not provide the desired volume percentages, those proportions were used as a starting point and then saline/oxygenated fluid was added appropriately until the desired volume percentages were obtained. The proportions of oxygenated fluid and saline used to get the desired volume percentage within ± 5 percent are listed in Table (6.2).

Table (6.2) - Experimentally Observed Mixing Proportions

Desired Volume %	Volume O2 (mL)	Volume Saline (mL)	Mass (g)	Total Volume (mL)	Actual Volume %
10	54	16	59.8	70	14.57
30	32	30	32.1	43	25.35
50	13	22	16.5	35	52.86
70	n/a	n/a	13.5	48	71.88
90	pre-made	pre-made	3.6	52	93.08

6.3 PARTICLE SIZE DISTRIBUTION

Particle size distribution can be determined by light obscuration using an AccuSizer, pictured in Figure 14. The AccuSizer analyzes a 10 microliter sample of a fluid and outputs mean particle diameter and the number percentage of particles exceeding 10 micrometers in diameter. Particles of diameter above that threshold increase the risk of LOMs becoming trapped within the vasculature and causing an oxygen gas embolus.



Figure (6.2) – AccuSizer 780A, Autodiluter by Particle Sizing Systems NIComp

The AccuSizer program gives an output plot distribution with a count of the number of particles of each diameter, an example of which is shown in Figure (6.3). As the plot shows, almost all of the particles are less than 10 micrometers in diameter, as needed, with the majority of the particles less than 5 micrometers in diameter.

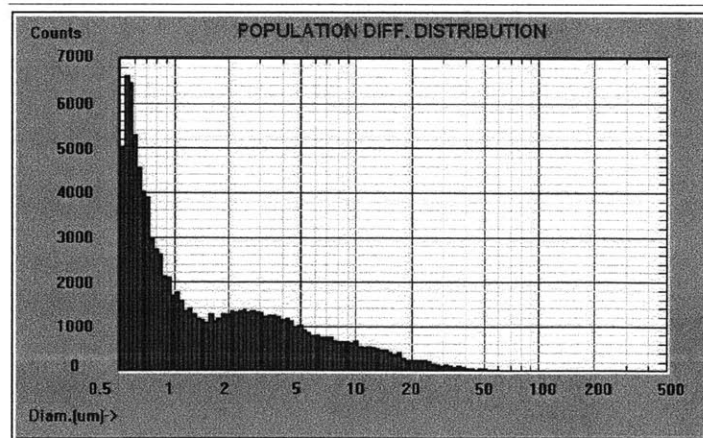


Figure (6.3) – Example of particle size distribution chart.

The mixtures made in the volume percentage experiments discussed earlier were used in the AccuSizer to determine the mean particle diameter for each sample. Table 3 lists the mean particle diameter for a range of volume percentages ranging from ten to ninety percent in twenty percent increments; the mean particle diameters listed are an average of three trials per volume percentage. All samples have a mean particle diameter of less than 10 micrometers, which is necessary in order to avoid air embolisms. There also seems to be a trend between volume percentage and particle size; it appears that as the gas by volume percentage increases, the mean particle diameter increases as well, with 52% being the only exception to that trend. This can be explained by fact that as the concentrated oxygenated fluid is diluted, the larger micro oxygen particles can be more easily released from the lipid molecules. As such, the more that the concentrated oxygenated fluid is diluted, the smaller the mean particle diameter.

Table (6.3) - Mean Particle Diameters

Volume Percent	Mean Particle Diameter (um)
13%	3.52
25%	4.49
52%	4.27
71%	5.74
92%	7.72

6.3.1 Microscopy (image for stratification)

Particle distribution can also be imaged using differential interference contrast and phase microscopy to ensure there is no stratification in the stored, concentrated fluid, as well as verify thorough mixing after the oxygenated fluid has been diluted with saline. There must be an even particle distribution using the current passive helical mixer in the system; if not, then it may need

to be replaced by a differently shaped passive mixer, like a 90-degree mixer, or an active mixer.

Figure (6.4) shows the Olympus microscope used here.

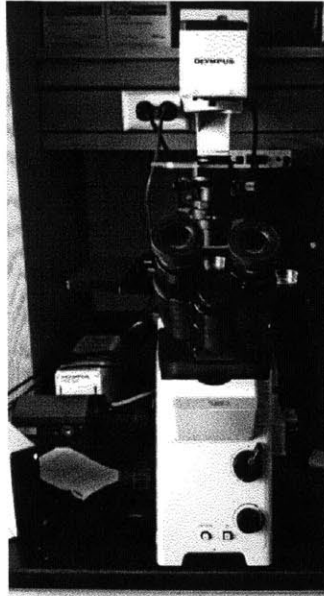


Figure (6.4) - Olympus Microscope

The images acquired using the Olympus microscope are displayed in Figure (6.5). These images confirm that there is an even particle distribution; there does not seem to be any sort of stratification or clustering of the microbubbles by size or shape. Since there is no stratification are slight hand agitation of the mixture, this is a good indication that the helical mixture may be sufficient for thorough mixing as well. Furthermore, there is no stratification of the 93 percent gas by volume sample which was pre-made by the doctor and stored overnight in a refrigerator, so storing the fluid for a day does not seem to result in size separation. These images also confirm the trend noted in the particle sizer data—as the volume percentage increases, the particle size increases with it. There is no image for the 13 percent gas by volume sample because it was so dilute that it was difficult to find any groups of bubbles.

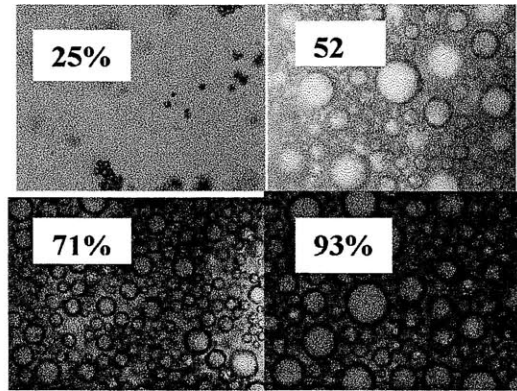


Figure (6.5) - Microscopy Images of Oxygenated Fluid

6.4 OXYGEN TENSION

Oxygen tension, pO_2 , is the partial pressure of a component of a gas dissolved in a fluid and is measured in mmHg [4]. Similarly, pCO_2 is the carbon dioxide partial pressure. Finally, pH quantifies the hydrogen ion concentration of a solution, where a pH of seven is a neutral solution. Oxygen tension, carbon dioxide tension, and pH can be calculated using a Clark electrode, pictured in Figure 18, by placing a 50 microliter aliquot of the fluid into a blood gas machine. The pH, PO_2 , and PCO_2 outputs from this machine can be used to further confirm repeatability of the fluid. If each concentration of the oxygenated fluid results in the same pH, pCO_2 , and pO_2 then the fluid and mixing methods are consistent. This test, however, is more difficult because any exposure to air can change the readings significantly.



Figure (6.6) - Blood Gas Machine

This blood gas machine was used for the five different samples, and the results of pH, pCO₂, and pO₂ are listed in Table (6.4). The data collected from the experiments will serve to create a model for ensuring that the correct volume of fluid at the desired concentration is delivered. As expected, the pH does not vary much from the neutral for any of the samples, and pCO₂ is much less than pO₂ for all samples as well.

Table (6.4) - pH, pCO₂, and pO₂ Results of Various Mixtures

Volume Percent	pH	pCO ₂ (mmHg)	pO ₂ (mmHg)
13%	6.72	4	592
25%	6.78	2	507
50%	6.63	9	749
70%	7.4	4	742
90%	7.44	5	346

6.5 RHEOMETRY

Rheometry measurements were obtained using a parallel plate geometry, shown in Figure (3.2), to find the steady state flow viscosity as stress is varied from 0.1 to 1,000 $\mu\text{N}\cdot\text{m}$. Rheologic profiles have been characterized for concentrations of 90 percent, 80 percent and 65 percent oxygen, shown in reference Figure (6.6).



Figure (6.7) – Rheometer

One goal of this research was to find the rheologic profile at approximately 70 percent gas by volume. The existing samples at 90 percent are used to calibrate the test machine. The data for 80 percent and 65 percent concentrations can be used to further verify the operations of the machine thus providing a closer region of calibration around 70 percent.

The doctor manufactured the 92 percent gas by volume sample, and this sample was then diluted using saline to create the 71, 52, 25, and 13 percent gas by volume samples. Figure (6.8) shows a graph of stress (Pa) versus viscosity (Pa-s), both on logarithmic scales, for all five samples. As expected, the 92 percent gas by volume sample has the highest viscosity, and viscosity decreases as the volume percentage decreases. In addition, the 13, 25, and 52 percent gas by volume samples all have linear relationships, which was expected since the 65 percent gas by volume reference profile shows a shift to a linear relationship as well. The 71 percent gas by volume sample, on the other hand, has three inflection points, as seen in the 90 and 80 percent gas by volume reference profiles.

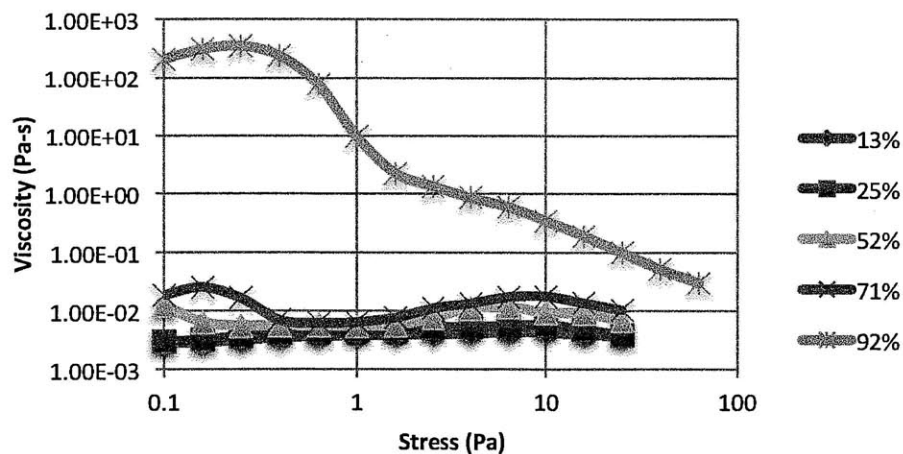


Figure (6.8) – Rheology profiles for a range of volume percentages.

However, there were a few discrepancies. First of all, each profile seems to have a viscosity smaller than the reference rheology profile by about two orders of magnitude. This difference is worrisome since even the 92 percent gas by volume sample, which was unchanged from when it was manufactured by the doctor varied greatly from the reference; this indicates that the fluid manufacturing method itself is not very repeatable. In addition, all of the reference profiles for the oxygenated fluid show a general trend of decreasing viscosity as the stress increases, but in the experimental data, the viscosities for 71, 52, 25, and 13 percent gas by volume remain relatively constant as the stress increases.

6.6 DEGRADATION

The concentrated oxygenated fluid is not always used immediately, so it is often stored in a refrigerator until it is needed. However, over time, the fluid seems to degrade and some oxygen microparticles are released into free gas. This causes some degradation of the fluid, and as was visually observed, this degradation occurs quickly over the first twenty-four hours, and then slows down significantly. If given more time, degradation of the concentrated form of the oxygenated fluid could be assessed by measuring the change in volume over a time and calculating the new volume percentage.

SUMMARY

7.1 SUMMARY

The oxygenated fluid pump described in this thesis has life saving implications. At its current state of development, it can be used as a tool to help characterize the fluid and perhaps run some basic animal testing. The design of the pump is extremely modular and has the capability of being easily modified for improvements. This device can successfully store and dispense one liter of fluid, mix the concentrated form of the oxygenated fluid with saline, maintain sterility, and preserve the fluid's properties. Further testing is required to tune the control system and ensure that the flow rate is accurate to $\pm 10\%$. The bubble trap needs to be debugged; it may necessary to create or purchase a bubble trap to handle the high infusion rate.

There are many smaller improvements that can be made to the device in the future. There are some zip ties that can be replaced with clips, the tubing and plumbing component layout can be improved, the part count can always be reduced, and a physical emergency "stop" button should be created. The oxygenated fluid pump also vibrates a fair amount due to the stepper motor. Vibrations in general however are not good and can cause bolts to loosen and can create noise in the signal from the load cell. It would be convenient to purchase a purge valve that would allow only air to flow out. The purging stage is critical for calibration and getting rid of any excess air

would allow for better calibration. It would also be beneficial to increase the portability of the device so that it can be easily used in emergency situations.

There are also a few improvements that can be made to the software. First and foremost, the calibration stage should be modified. The control system is extremely dependent on the calibration. If the calibration goes wrong, then the code will not behave in the manner that it is suppose to since the control system is consistently referring to the calibration data. An easy fix for this would be to run the calibration multiple times with different batches of fluid. This way, a model can be developed for the fluid an calibration would no longer be necessary. This would potentially get rid of the calibration phase altogether or it can change the way it behaves. A calibration can still be run on the fluid, but instead of using it in the controls, the fluid calibration would just be compared to a fluid model. If the particular batch of fluid deviates too much from the model, then it is a bad batch, if it is similar, then pumping can be initiated.

Much testing needs to be done to determine the reliability of the pump. A good strategy would be to mix small amount of the oxygenated fluid with either water or saline, slowly increasing the concentration of oxygenated fluid until the 90% concentration is achieved. This way, if something goes wrong, determining the cause of error will be easier since the viscosity of the fluid will be increasing slowly. There are also a few caveats to keep in mind. The load cell signal is very dependent on the electronics being properly grounded. A computer is necessary to run the labview code and to control the pump. The computer needs to be plugged in to an outlet in order for the system to be properly grounded. If it is not, the signal will be extremely noisy and the pump will not function correctly.

REFERENCES

[1] Personal communications with Dr. John Kheir, October through December 2011.

[2] “The Belmont Rapid Infuser”. Belmont Instrument Corporation.
<http://www.belmontinstrument.com/products/rapidInfuser>

[3] “Tool for knee surgery will help rebuild ACL”. Center for Integration of Medicine & Innovative Technology.
<http://www.cimit.org/about-stories-kneesurgery.html>

A

MOTOR SPECIFICATIONS

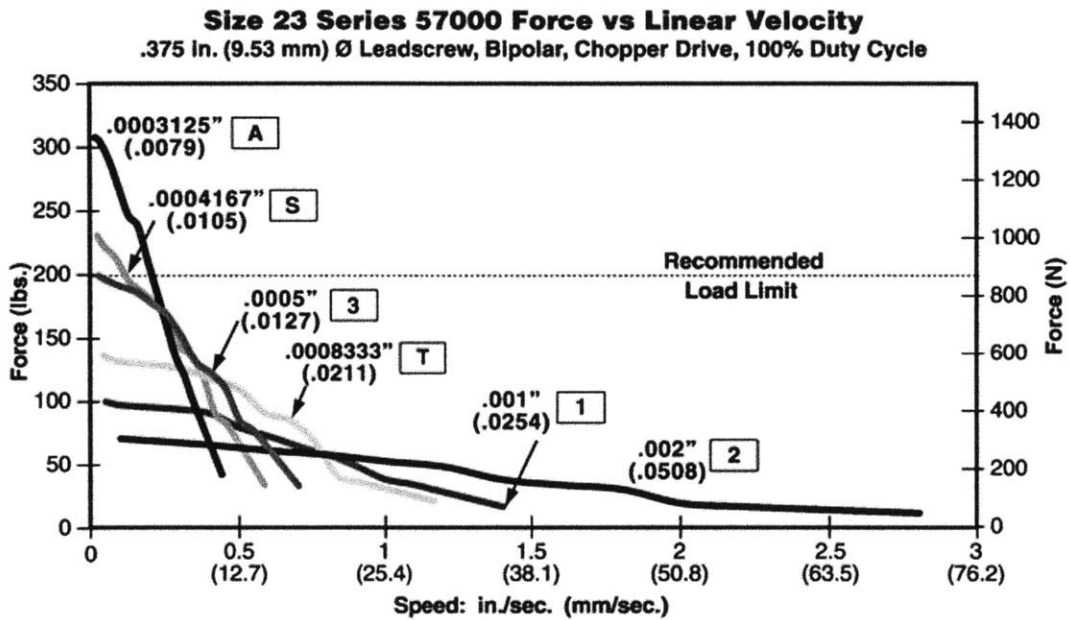


Figure (A.1) – Force vs. Speed curve for stepper motor.

This torque curve is from the Haydon Kerk website. The motor used in this application is the one labeled as 1. It has a step size of 0.001" and a maximum force of 100 pounds.

APPENDIX

B

BUDGET

Table (B.1) – This table outlines all the expenditures of the oxygenated fluid pump.

Category	Item	Price	Notes
Peristaltic			Thomas Scientific
Structural Materials	Aluminum Inch T-Slotted Framing System 90 Degree Bracket, Single, 2-Hole, 1/2" x 1/2" x 1/2"	\$19.90	McMaster-Carr (47065T223) (5)
	Aluminum Inch T-Slotted Framing System Plate, Single, 2-Hole, for 1" Extrusion	\$45.80	McMaster-Carr (47065T141) (10)
	Std Zinc-Plated STL End-Feed Fastener, for 1" Aluminum Inch T-Slotted Framing System	\$4.60	McMaster-Carr (47065T142) (2)
	Compact Head End-Feed Fastener, for 1" Aluminum Inch T-Slotted Framing System	\$7.40	McMaster-Carr (47065T139) (4)
	Nylon Thrust Bearing for 1/4" Shaft Dia, 5/8" OD, 1/16" Thickness	\$4.80	McMaster-Carr (2797T1) (4)
	Multipurpose Aluminum (Alloy 6061) .125" Thick, 12" X 12"	\$28.02	McMaster-Carr (89015K18)
	Waterjet new rail and outrigger	\$51.00	MIT Hobby Shop
Measurement	Graduation Cylinders, 500mL (x2)	\$30.00	VWR
Electronics	Amplifier (DMD-4059-DC)	\$379.00	Omega
Plumbing Materials	1/2" fullport ball valve FPT 400PSI	\$7.32	Home Depot
	1/2 quick connect x 1/2 MIP	\$2.98	Home Depot
	5/8" OD x 1/2" ID x 10' vinyl tube	\$4.38	Home Depot
	1/2" OD x 3/8" ID x 20' vinyl tube	\$6.90	Home Depot
	2x1 2Nylon BarbxMIP	\$2.84	Home Depot
	White Polypropylene Push-to-Conn Tube Fitting Reducing Coupling for 1/2" x 1/2"	\$9.46	McMaster-Carr (9087K66) (2)
	Moisture-Resistant Acetal Push-to-Connect Wye for 1/2" Tube OD, Gray	\$15.08	McMaster-Carr (51055K418) (2)
	Pressure Steel Thrd Fitting 3/4"-16 UNF-2A X 1/2" NPTF Male Adapter	\$14.72	McMaster-Carr (50925K332) (2)
	Type 316 Stainless Steel Hex Nut 3/4"-16 Thread Size, 1-1/8" Width, 41/64" H	\$11.72	McMaster-Carr (94804A365) (1)
	O-Rings (pack of 20) width 3/16" OD 3.27" ID 2.85"	\$11.23	McMaster-Carr
Bellows Cap	\$312.00	3D Printsmith LLC, Durus material Billed to Biomedical Modeling, Inc.	
Other	6 outlet power strip, 3 ft cord	\$3.72	Home Depot
Total Spent		\$921.87	

C

ANNOTATED LABVIEW CODE

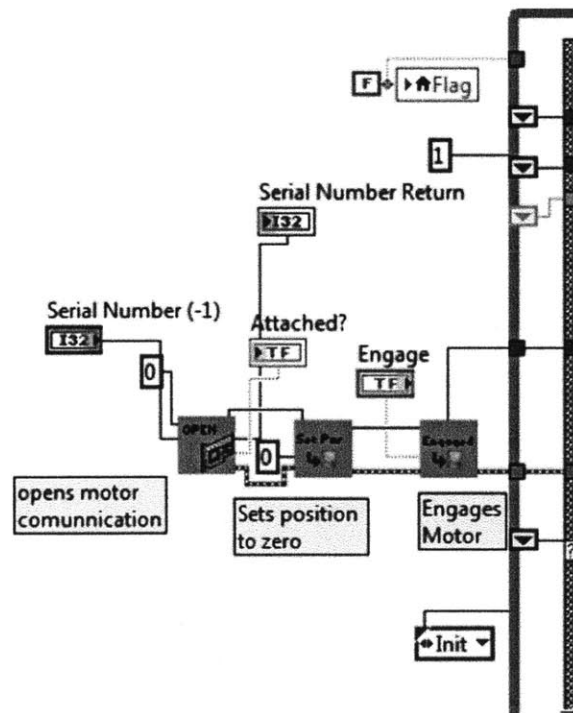


Figure (C.1) – This section of code opens communication with the motor.

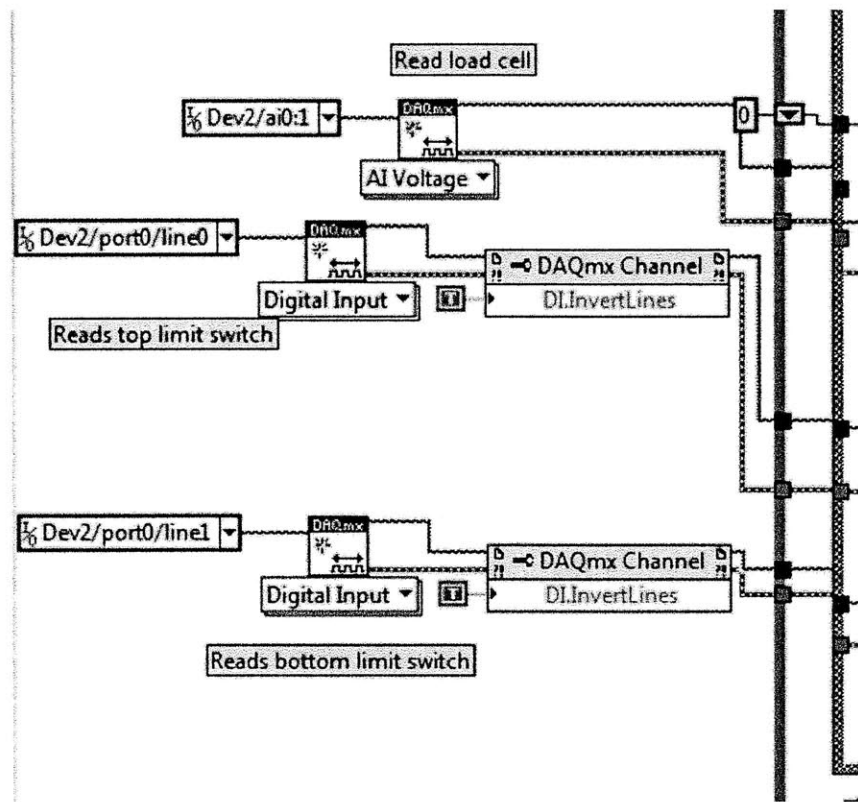


Figure (C.2) – This section of code opens communication with the limit switches and load cell.

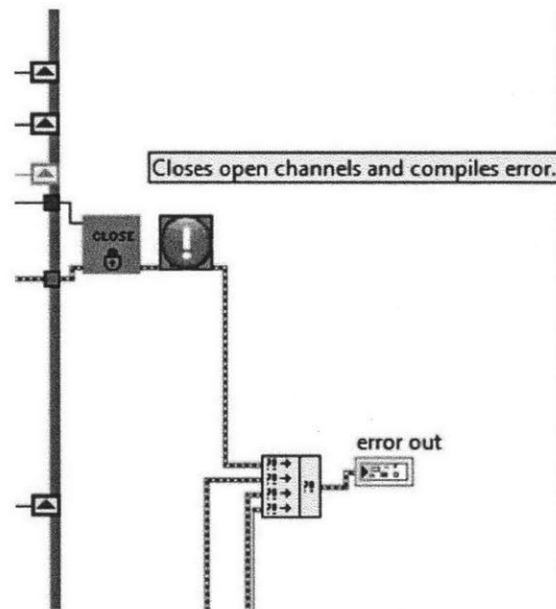


Figure (C.3) – This block closes communication with the load cell, limit switches and motor.

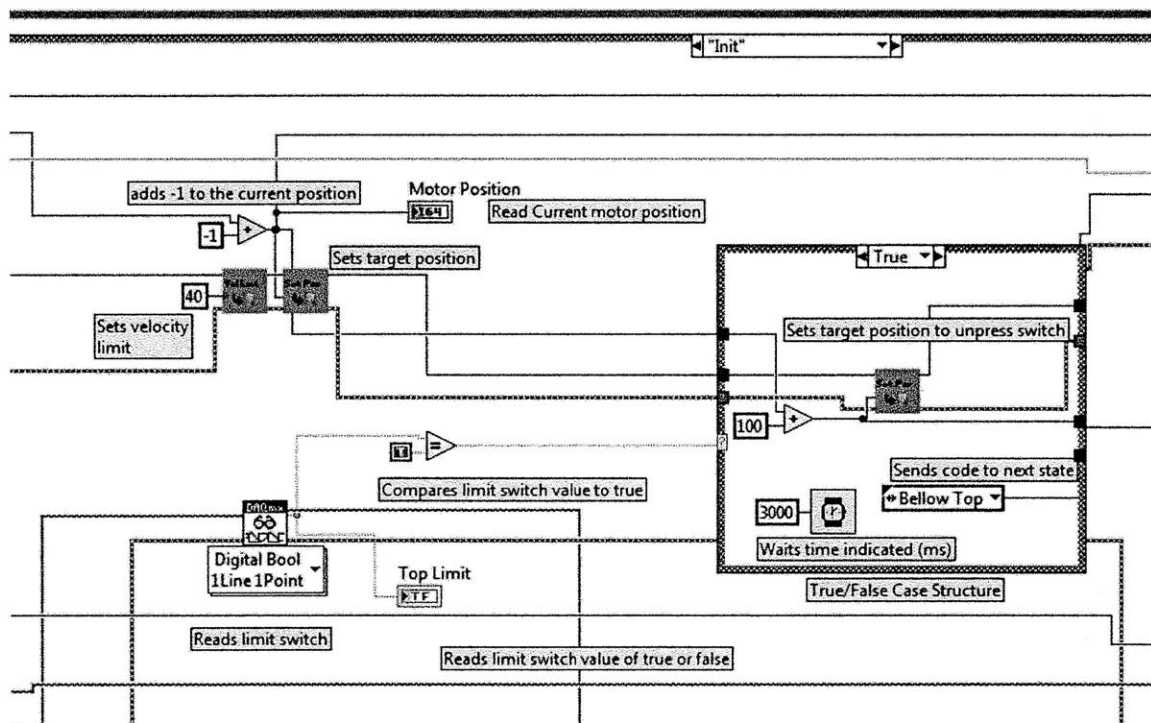


Figure (C.4) – Initialize Phase

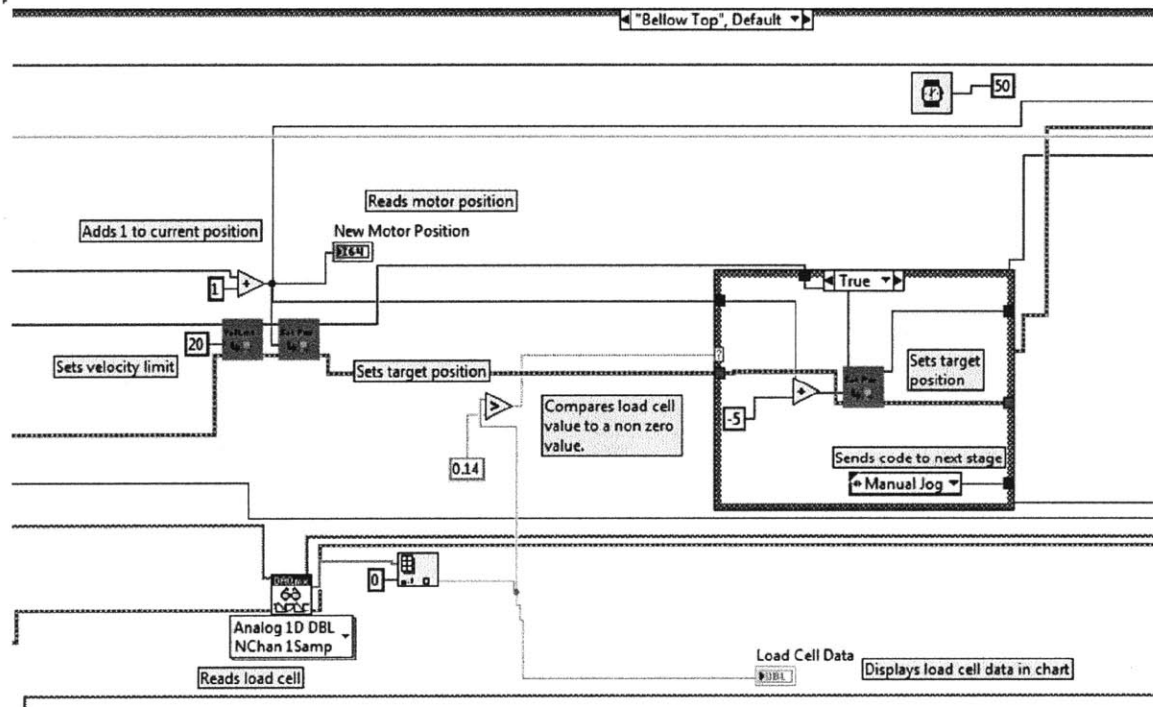


Figure (C.5) – Bellows Top Phase

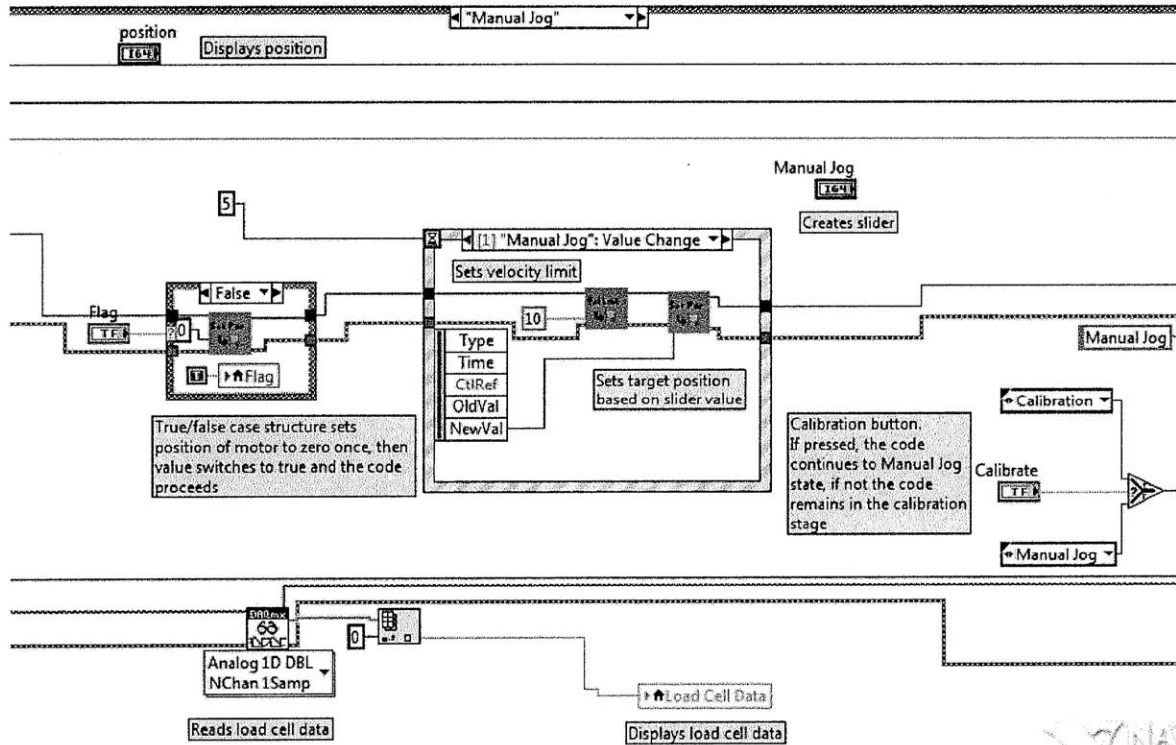


Figure (C.6) – Manual Jog Phase

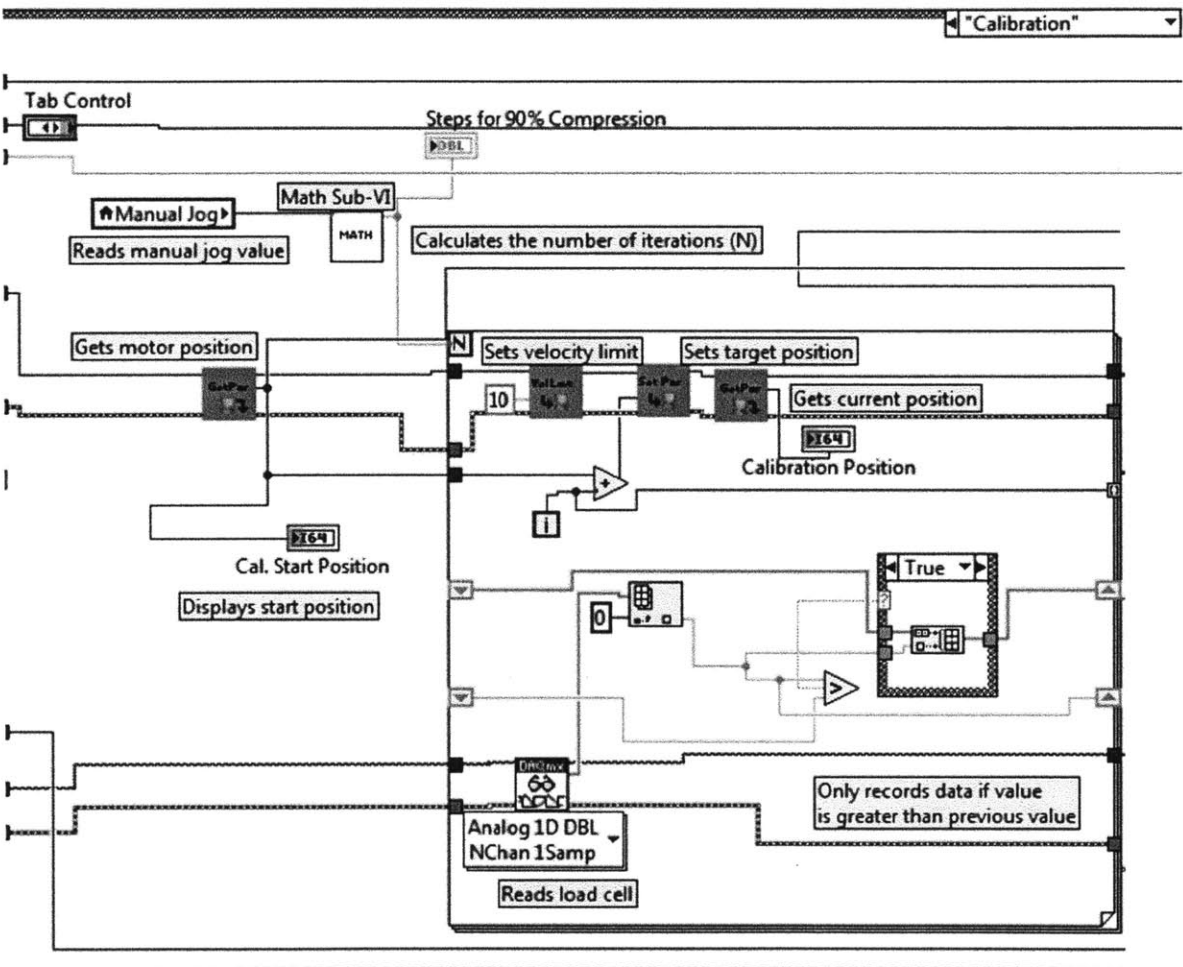


Figure (C.7) – Calibration Phase (1 of 2)

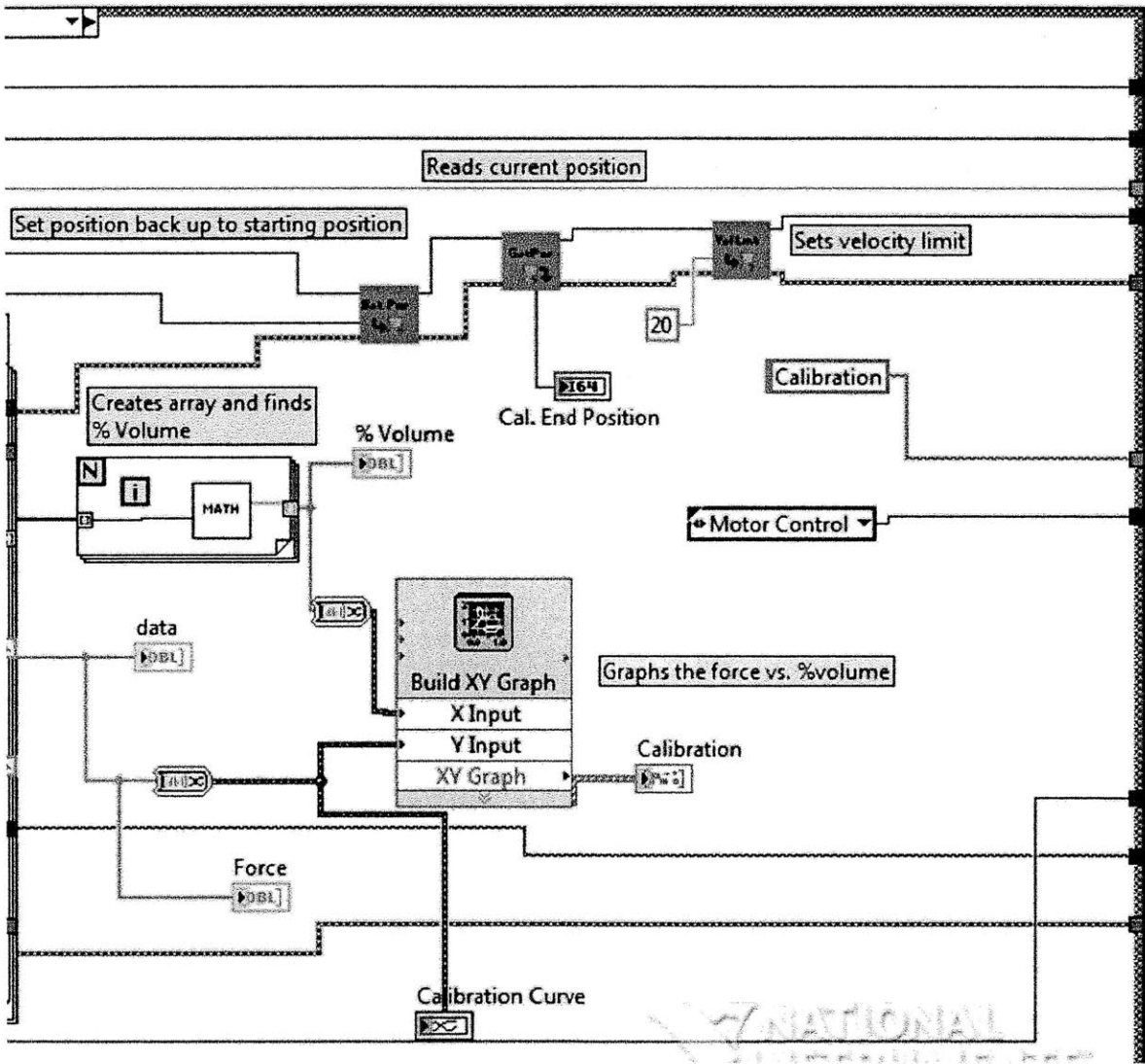


Figure (C.8) – Calibration Phase (2 of 2)

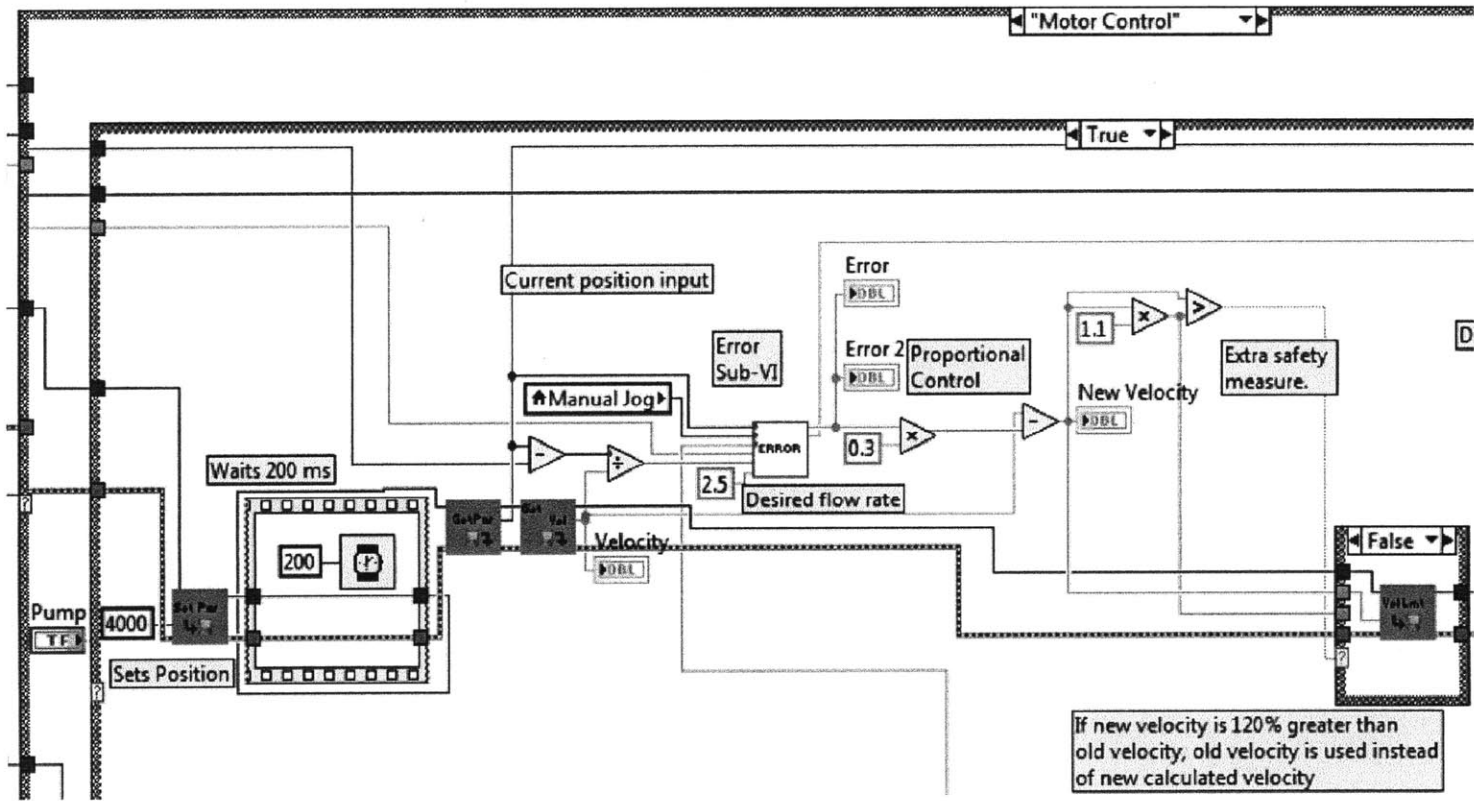


Figure (C.9) – Motor Control (1 of 3)

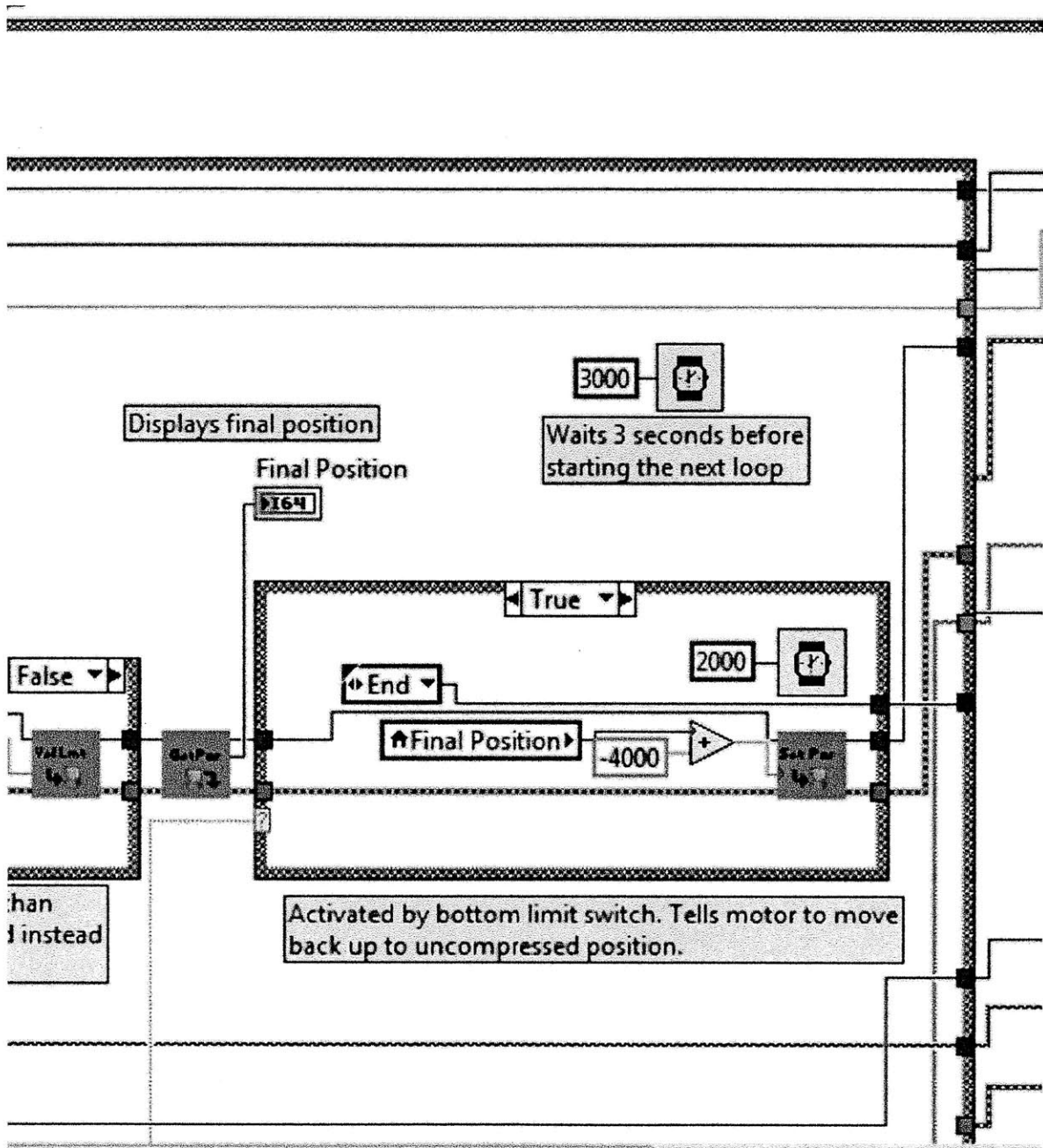


Figure (C.10) – Motor Control (2 of 3)

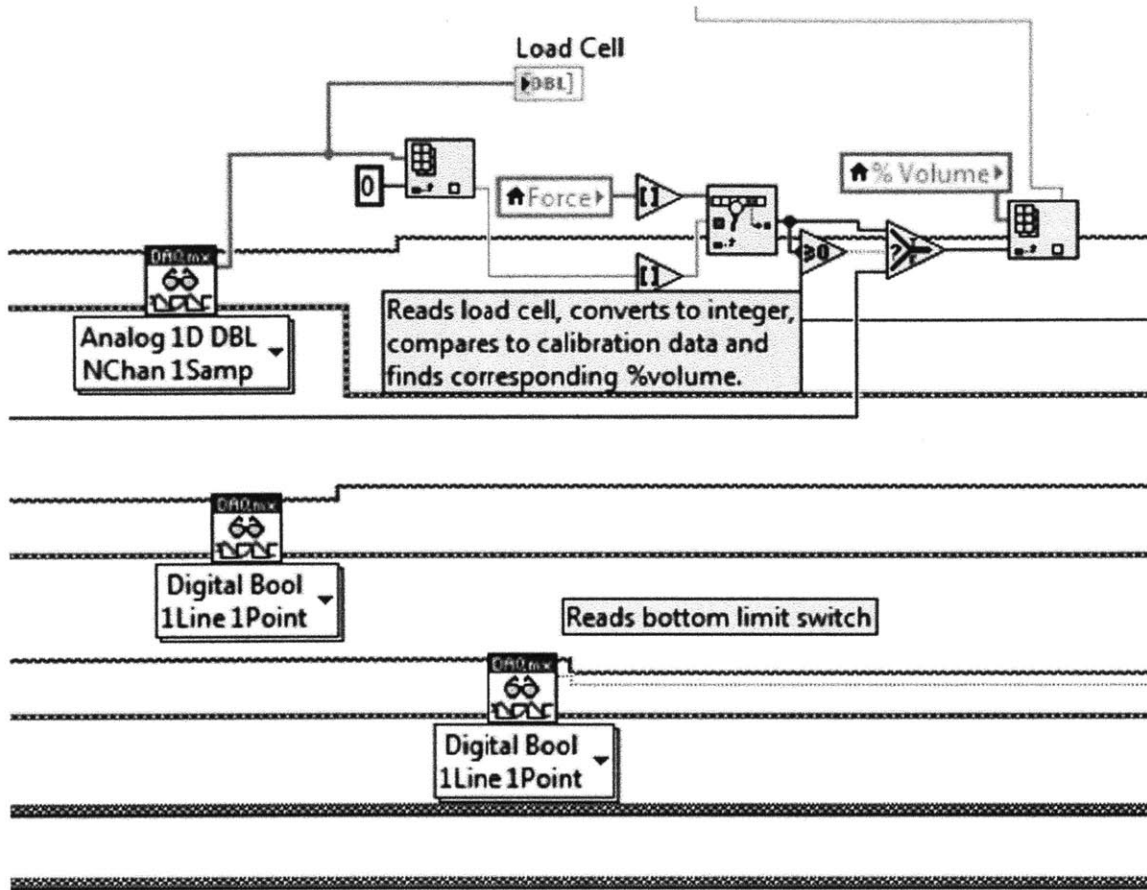


Figure (C.11) – Motor Control (3 of 3)

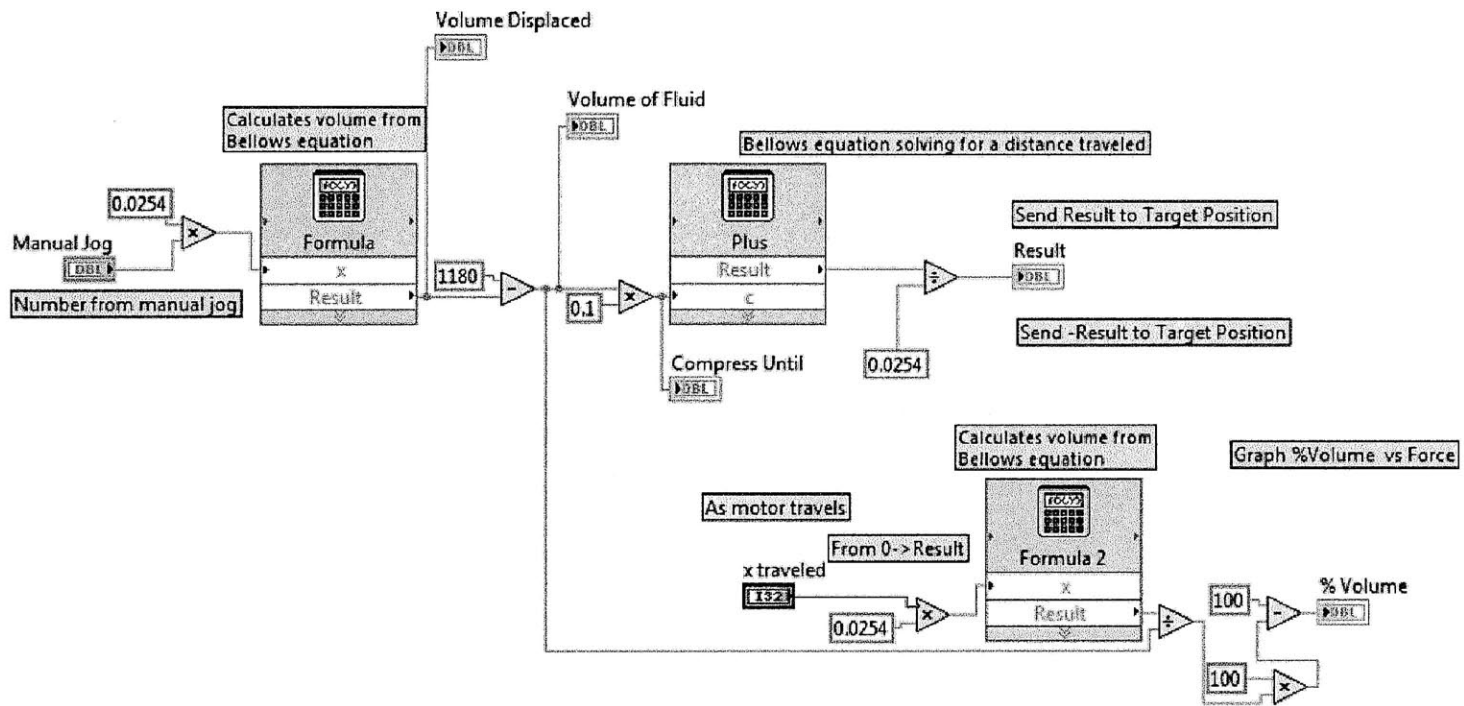


Figure (C.12) – Math Sub-VI, calculates %volume and distance to 90% compression

Motor position including Manual Purge after time dt

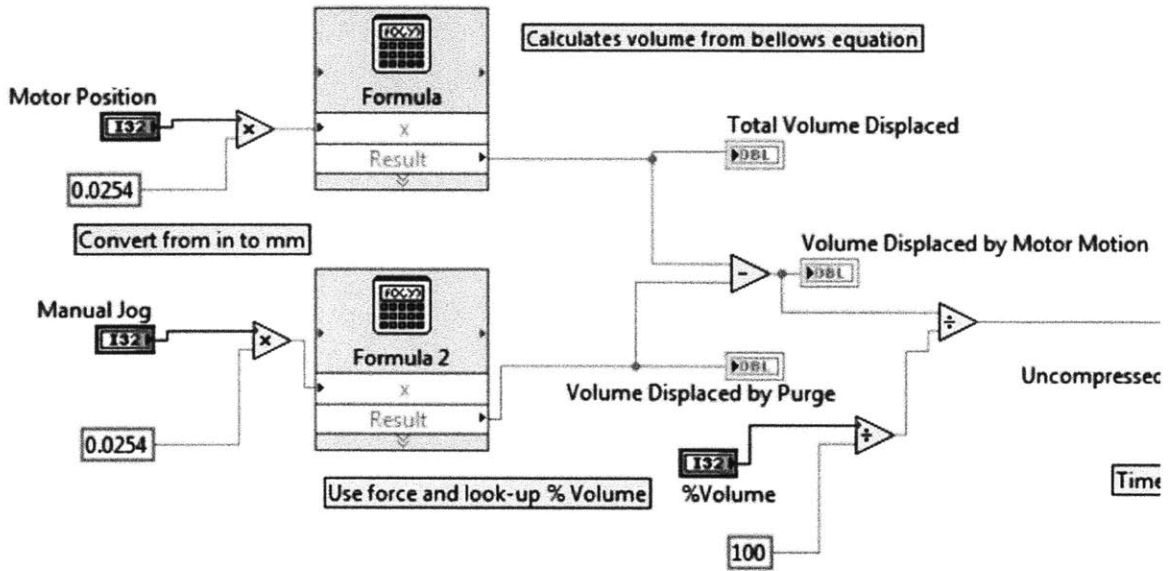


Figure (C.13) – Error Calculation Sub-VI (1 of 2)

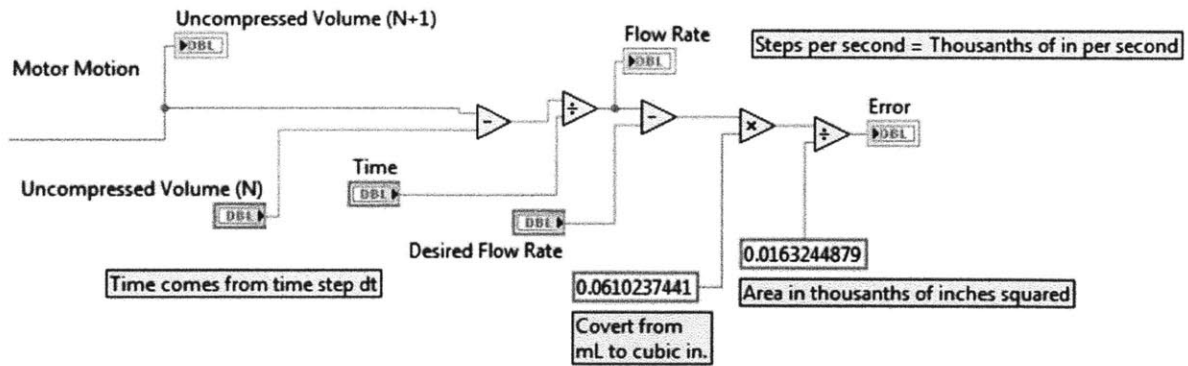


Figure (C.14) – Error Calculation Sub-VI (2 of 2)