**Syringe Pump Final Report**

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**Executive Summary:**

Our team was asked to design a syringe pump that would deliver fluid at a controlled flow rate to cells in a microfluidic device. The design process of our syringe pump proved to be a very dynamic one. The beginning research of both microfluidic devices and existing syringe pumps helped our team get an idea of ways we could implement existing aspects that work into our design. There were many existing devices that resembled the one that we were asked to make closely; however, due to our resources as students, we had to be a bit more creative in figuring out how to afford and assemble each component to the best of our abilities. Developing customer requirements was a huge step in the process of understanding what exactly you as our customer wanted to see delivered in our syringe pump. The main requirements of our pump were that it was able to deliver accurate shear stress values so that they could mimic those found in true physiology, that it was able to deliver an accurate flow rate to the device, that it was easily usable, and that it was compact to both fit in a desired location and have ease of mobility when needed to be moved to or from that location. Next, it was our job as the engineers to turn those requirements into quantitative engineering specifications that our device needed to meet via testing of the device once the prototype was finished. Once we determined what numbers needed to be hit to quantify the requirements set by you, we were able to create a network diagram of tasks in order to organize the design, manufacturing, and testing processes that we had ahead.

Our design process then became a series of brainstorming via tools like a conjoint analysis, morphology, and Pugh matrices. We did these exercises in order to compile a multitude of ideas for each component of the pump to determine which combination of these ideas would produce the optimal pump that is attractive to the user and does the best job at meeting the customer specifications. We determined the main functions of our pump were inputting flow rate parameters on the interface, having a power source for the pushing mechanism, the physical pushing mechanism, and lastly the mechanism through which the fluid would be delivered into the tube. Ultimately, through the many exercises as well as iterations due to a multitude of realizations down the road, we settled upon using a stepper motor linear actuator for the pushing mechanism and a screen with buttons for the input from the user, powered by a 24 V DC Power Supply and connected by a needle attachment to the syringe. Next came acquiring the materials and aspects of the pump that were to be purchased from a manufacturer as well as designing the aspects that we were going to manufacture ourselves. The primary component of our design that we purchased was the FUYU stepper motor linear actuator, to which we programmed electrically and designed adapters to fit onto. Our electrical programming revolved around the Arduino UNO and the Sketch coding software. The chassis was our last component to design, and its main purpose was to keep the user safe from any potential harm from the pump and

protect the pump from any water or other wear. When we had performed the Hazard Safety Assessment, we determined a lot of the risk involved the user having their hands in the pinch points as well as having the device fall on the user, both of which were mitigated by having a chassis that covered the pinch point and made the device more compact and mobile. Once we had those components designed, we determined how we would both manufacture and assemble the final prototype. These plans were surely dynamic as we changed materials and found new ways to better manufacture each piece. Critical changes included changing the chassis material from acrylic to polycarbonate, and thus changing the manufacturing process from laser jetting to water jetting to using a variety of saws to cut the pieces. Another critical change came after having manufactured the pusher block adapter, as we were sent back to the design process when the adapter did not perform the way we wanted it to. Additionally, the electrical side of our design manufacturing had to be iterated multiple times as we determined what was feasible and still effective for inputting the parameters. Our design changed from a 4 x 4 keypad to two buttons, one increasing the flow rate value and one decreasing the value. Once the prototype had been built, it was time to verify that we had made a device that met the customer specifications. We created protocols for how we would test these specifications and executed each of the four, the most time-consuming ones being the flow rate and shear stress tests.

Our testing plans for shear stress included both an analytical COMSOL simulation through the solid model of the microfluidic device as well as physical testing of the velocity of the particles moving via the LabSmith Micro Particle Image Velocimetry microscope. The physical testing was to verify that our analytical model accurately displayed what velocity and thus shear stresses the cells in our microfluidic devices would be experiencing. Next, we tested flow rate via running water through our pump at specified flow rates for a given period of time, measuring the mass acquired on a sensitive scale to back-calculate what flow rate was actually being delivered. Additionally, we used a gauge to measure the displacement of our pusher block over a specified time to first ensure that the correct speed was being programmed to the motor. In terms of surface area testing, we simply used a ruler to measure the dimensions of the bottom of our chassis to verify it would fit in the desired location in the lab. Lastly, our ease of use testing included simply numbering the steps in the operations manual.

Ultimately, our data showed that we did in fact create a pump that received an input and delivered a controllable flow rate and shear stress to the cells in the microfluidic devices, all while being compact and easily usable. After inputting a flow rate of 28.8 ml/hr, we measured the delivered flow rate to be 25.5 ml/hr, which was within our target percent error range of 15%. For shear stress, when entering a flow rate of 75.8 uL/hr, our physical testing showed a particle velocity of 295.6 um/s and our COMSOL velocity showed one of 358.91 um/s, putting these within range of our 20% error goal. We measured the bottom surface area of our pump to be 431.85 cm^2,

which was well within our specification of 695 cm^2. Lastly, we measured 5 steps to program our device, which was our target specification. There were surely limitations to our data, as when flow rate decreased to smaller and smaller values it was increasingly harder to acquire data, and then additionally extremely difficult to have that data be accurate. Thus, at the flow rate of 0.76 uL/hr, which is the flow rate at which the pump will typically be used at, both the shear stress and flow rate specifications were not met via our testing. There are a multitude of reasons why our data may have been skewed, and we have plans for future testing to discover where errors might be introduced in our pump. Overall, our team learned much about the design process and grew as engineers while designing this syringe pump.

## Statement of Work

### Abstract of Statement of Work

This statement of work will include a brief introduction of our project, the syringe pump, as well as a background on relevant information that pertains to current technologies, what our sponsor already has in place, and standards or codes in place that will affect our design. This document also includes the objectives of our project and our strategies of project management.

### Introduction for Statement of Work

Our syringe pump project aims to create a device that has the ability to supply fluid into a microfluidic device with flow rates that are able to be controlled to mimic flow velocities found in natural physiology. The main stakeholders of our project are the people of the Microphysiological Systems Lab, specifically under the leadership of Professor Heylman, who will be using this project for their research. The goal of our project is to create CFD simulations of flow through a microfluidic device that would be used in the lab to determine the flow rates needed to create physiological shear stress. From there, the goal is to create a functional prototype of a syringe pump that can deliver said controlled flow rates in order to enhance research by making the system more biomimetic. We have set additional goals pertaining to meeting specific engineering speculations based on the requirements you have provided us, including controlling the shear stress, the amount of fluid the pump dispenses, the parts of the cells that stay intact through shear stress, as well as ease of use and manufacturing.

### Background

The problem at hand is the need for a syringe pump to deliver fluids at constant, controllable flow rates to microfluidic devices in order to impart a shear stress on the cells in the device that mimics what would be felt by the cells in their true physiology. We understand that the MPS Lab is in need of their own syringe pump system in order

to conduct research regarding drug delivery in these microfluidic devices. We aim to improve upon the syringe pump that the lab already borrows. After discussing with the grad students who are the primary users of the current system, they have expressed that the two of the main concerns of the current pump are the unintentional removal of ECM cells and the size and surface area of the pump. We understand that the cells inside the microfluidic devices for which we are designing the syringe pump are endothelial cells, which are the most mechanically sensitive cells and thus something we must take into account when calculating flow rate for required shear stress.

Microfluidic syringe pumps are used to deliver fluids on small scales in a variety of fields (i.e. The medical industry to administer drugs, manufacturing of computer chips, microfluidic research). After researching syringe pumps currently available that deliver fluids to microfluidic devices, we have noticed patterns in how they mostly work. The syringe pump's mechanical components are simple; an engine drives a piston linearly, causing fluid to be pumped out of the syringe. A computer is used to control the specific amount of fluid that is administered by the syringe, allowing the user a high level of accuracy. The precision of the syringe pump is decided by the tolerance of the mechanical components of the pump. Consequently, the focus of our design will be on the mechanical components of the syringe pump and not building the software that controls the piston. Companies such as Harvard Apparatus and Celix already have high performance microfluidic syringe pumps- these companies will be the standard that we will surpass with our design. Harvard Apparatus offers the PHD-Ultra 70-3007 High Performance Syringe Pump, which offers a continuous max flow rate of 216uL/min and a min flow rate of 1.56pL/min with an accepted tolerance of ± 0.25%. Celix's ExiGo pump offers a continuous max flow rate of 13mL/min and minimum 10nL/min. The application of our syringe pump won't necessitate such a wide range of flow rates, but we aim to provide any necessary flow rate with tolerances that match industry standards.

The deliverables for these systems pertain to microfluidic devices in addition to extracellular matrix gels that contain human cells. The cells require both nutrients delivered as well as waste removed in order to maintain efficiency. These provision and removal processes are transported through the flow of nutrient media, both of which travel through the microfluidic channels of the device. Biomimicry can be improved by using the flows of media to impart shear stress on the cells present.

Obviously, there are many uses for and designs of syringe pumps already patented and on the market. The table below summarizes our findings on devices similar to the one we aim to create, as well as their patents. Also in the table is how we might use ideas from the patents appropriately.

## Intellectual Property Assessment

**Table 1: Related Devices and Patents**

| Name of Patent | Patent ID | Description of Device/ Patent |
|---|---|---|

| | | |
|---|---|---|
| Microfluidic Delivery Device | US20150025461A1 | • A microfluidic delivery device that pumps a predetermined volume of liquid consisting of a channel that connects inlet and outlet, a moveable member that controls positive pressure versus negative pressure, and a cavity capable of accepting fluid from the channel when the member is moved to fill position stop.<br>• Our design would be changed to focus on the flow rate of the liquid that enters the device rather than the predetermined volume. Can also file for a license if we want to use the inlet and outlet ideas. |

| | | |
|---|---|---|
| Syringe Pump | US7635349B2 | • A syringe pump involves a motorized rotating leadscrew that pushes a plunger head retainer through the barrel of the syringe resulting in the dispensary of the medication within the syringe to the patient. The head retainer includes a pressure sensor which is able to measure the applied force and has the ability to detect an obstacle restricting medication flow.<br>• Our design would be similar but the syringe pump is delivered to a microfluidic device rather than to the patient. |

| | | |
|---|---|---|
| Generating a Fluid Stream in a Microfluidic Device | US20190204209A1 | • A fluid handling and delivery system that generates fluid streams in the path of a microfluidic device. The patent defines the apparatus as including a plurality of receptacles defining interior space that maintains amount of gas pressure, gas inlet, a plurality of variable volume containers, and a plurality of fluid outlet elements.<br>• We might file for a license or operate at risk in order to use these ideas to maintain pressure to stabilize desired flow rate in the microfluidic device |

| | | |
|---|---|---|
| MICROFLUIDIC qRT-PCR ANALYSIS OF SINGLE CELLS<br>(Patent Application) | 20170283859 | ● A microfluidic device for single cell gene expression analysis comprising: a cell inlet configured to receive a fluid containing a plurality of cells; and one or more analysis units coupled to the cell inlet, wherein each of the one or more analysis units comprise: a cell trap configured to trap a single cell from the plurality of cells; a reaction chamber coupled to the cell trap<br>● We can potentially use these ideas but change the design to focus not on delivering cells but delivering fluid to the cells that will be in our microfluidic device |

| | | |
|---|---|---|
| PERISTALTIC PUMP (Patent Application) | 20210304864 | • The peristaltic pump includes a plunger, inlet and outlet valves, a spring, and an actuator. The plunger actuates toward and away from a tube, the inlet valve is upstream of the plunger, the outlet valve is downstream of the plunger, the spring biases the plunger toward the tube, and the actuator mechanically engages and disengages from the plunger<br>• Our design change would be to apply these ideas to the syringe pump attached to a microfluidic device rather than a tube, thus tweaking the size parameters and the design to do so |

| | | |
|---|---|---|
| MICROFLUIDIC PRESSURE SENSOR (Patent Application) | 20210190616 | • A microfluidic pressure sensor may include a reference chamber, a sensed volume, a microfluidic channel connecting an interior of the reference chamber to an interior of the sensed volume, a volume of liquid contained and movable within the microfluidic channel while occluding the microfluidic channel and a sensor to output signals indicating positioning of the volume of liquid along the microfluidic channel<br>• We could apply these ideas to have a pressure sensor in the syringe pump rather than on the microfluidic device, so that can be our design change |

Additionally, we are aware that there is much information already published that we can use to design our syringe pump. Below is a table that details relevant technical literature that may be relevant in designing our project.

**Table 2: Relevant Technical Literature**

| Title | Author(s) |
|---|---|
| “A passive pumping method for microfluidic devices” | Glenn M. Walker & David J Beebe |
| “Syringe-pump-induced fluctuation in all-aqueous microfluidic system implications for flow rate accuracy” | Zida Li, Sze Yi Mak, Alban Sauret, & Ho Cheung Shum |
| “Pumps for microfluidic cell culture” | Chang Kyu Byun,Kameel Abi-Samra,Yoon-Kyoung Cho,& Shuichi Takayama |
| “A Simple Approach for Controlling an Open-Source Syringe Pump” | Fatih Akkoyun & Adem Özecik |
| “Three-dimensional Printing of Thermoplastic Materials to Create Automated Syringe Pumps with Feedback Control for Microfluidic Applications” | Ming-Cheng Chen, John R. Lake, Keith C. Heyde, & Warren C. Ruder |

Lastly, we have researched industry codes, standards, and regulations relevant to our project that may have an effect on how we go about creating the syringe pump. Our findings showed a lot of regulations for syringe pumps that were to be used for infusion into a human, but because our syringe pump is being used for research in microfluidic devices rather than real drug administration, no codes or standards were found that will restrict our particular device.

**Objectives (SOW)**

The objective of our project is to create a functional syringe pump that will deliver media to microfluidic devices in a controlled manner. We also strive to address the

concerns expressed by the current users of the syringe pumps in the Cal Poly labs. Our project is needed for research regarding the effect of certain drugs on human cells, as the more accurate the stresses imparted on the cells in the microfluidic device, the more valuable and accurate the results of the research will be. Our project explicitly includes the CFD simulations we will run in order to find the required flow rate to impart physiological shear stress on the cells, as well as a working syringe pump that is able to deliver these flow rates to the microfluidic devices. Our project will not include any additional work on the microfluidic devices themselves, nor does it include the environment which the microfluidic devices are to be stored in. The Indications for Use for our project which describes the full scope of the intended use for our syringe pump can be found in this document under the Indications For Use heading. Additionally, we have completed a Quality Function Deployment Analysis, and our work from that can be summarized in the table below. It is also organized in the House of Quality we created for the project (see Customer Requirements).

**Project Management**

The main deliverables of our project include the CFD simulations as well as the functional prototype of the syringe pump. The Network Diagram was created to ensure the team maintains stays on track of all of our deliverables. The network diagram, with the critical path, can be seen in the figures below. As our design process progresses, this living document will change to reflect the evolving needs of our project (see Network Diagram).

*Network Diagram*

To begin, we have the network diagram which organizes the assignments and deliverables that we need to complete, as well as dependencies between the tasks that lead us to a critical path. This will keep us on track to the timeline we have set to develop the syringe pump with an appropriate and well-structured design process. Note that between the conceptual design review and the next status update, additional COMSOL simulations will be conducted (item 15)

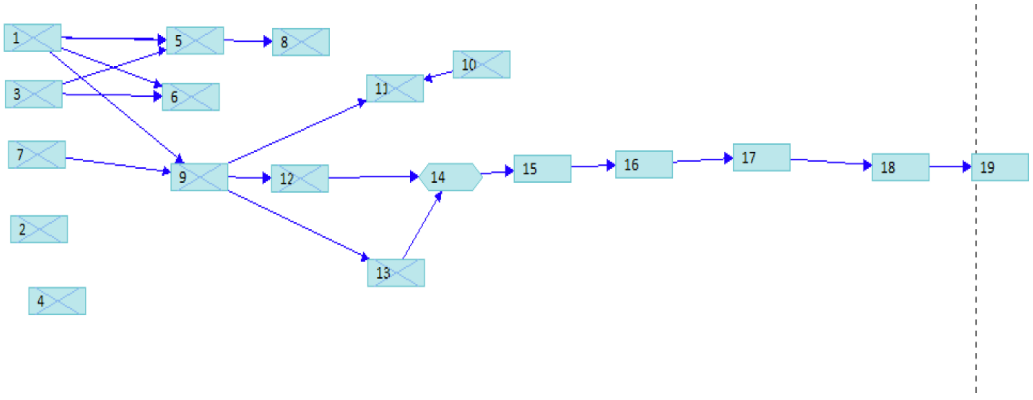


*Figure 1: This is an image of the Network Diagram, including the first half of all of the tasks we have to complete to reach our end goal.*

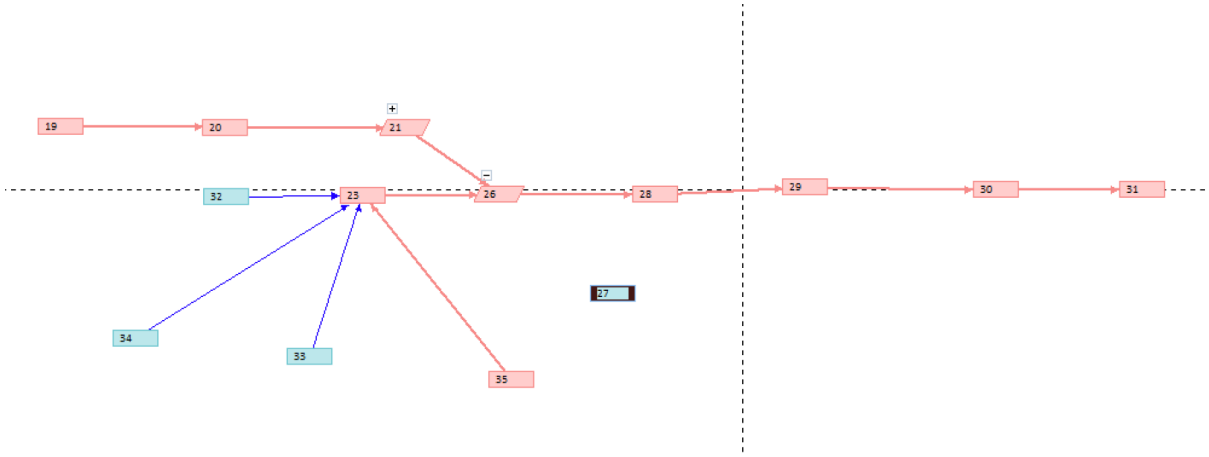



*Figure 2: This is an image of the critical path from our Network Diagram.*

| | Task Mode | Task Name | Duration | Start | Finish | Predecessors |
|---|---|---|---|---|---|---|
| 1 | | Complete House of Quality | 5 days | Wed 9/29/21 | Tue 10/5/21 | |
| 2 | | Network Diagram | 2 days | Mon 10/4/21 | Tue 10/5/21 | |
| 3 | | Meet with Grad Students | 4 days | Wed 9/29/21 | Mon 10/4/21 | |
| 4 | | IFU | 2 days | Tue 10/5/21 | Wed 10/6/21 | |
| 5 | | Develop Budget | 2 days | Thu 10/7/21 | Fri 10/8/21 | 1,3 |
| 6 | | Finish Statement of Work | 3 days | Wed 10/6/21 | Fri 10/8/21 | 1,3 |
| 7 | | Conjoint Analysis ANOVA | 4 days | Wed 10/6/21 | Mon 10/11/21 | |
| 8 | | Project Plan Meeting Slides | 4 days | Wed 10/6/21 | Mon 10/11/21 | 1 |
| 9 | | Morphology and Conc | 5 days | Tue 10/12/21 | Mon 10/18/21 | 1,7 |
| 10 | | Pugh Chart | 3 days | Mon 10/18/2 | Wed 10/20/2 | |
| 11 | | Status Update Memo #1 | 3 days | Thu 10/21/21 | Mon 10/25/21 | 9,10 |
| 12 | | Conceptual Model | 7 days | Tue 10/19/2 | Wed 10/27/2 | 9 |

| | Task Mode | Task Name | Duration | Start | Finish | Predecessors |
|---|---|---|---|---|---|---|
| 10 | | Pugh Chart | 3 days | Mon 10/18/2 | Wed 10/20/2 | |
| 11 | | Status Update Memo #1 | 3 days | Thu 10/21/21 | Mon 10/25/21 | 9,10 |
| 12 | | Conceptual Model | 7 days | Tue 10/19/2: | Wed 10/27/2 | 9 |
| 13 | | FMEA | 7 days | Tue 10/19/2: | Wed 10/27/2 | 9 |
| 14 | | Conceptual Design Prototype REVIEW | 3 days | Thu 10/28/21 | Mon 11/1/21 | 12,13 |
| 15 | | Status Update Memo #2 | 5 days | Tue 11/2/21 | Mon 11/8/21 | 14 |
| 16 | | Status Upate Memo #3 | 5 days | Tue 11/9/21 | Mon 11/15/21 | 15 |
| 17 | | Hazard Risk Assessment | 3 days | Thu 11/11/21 | Mon 11/15/21 | |
| 18 | | Critical Design Review Report and Presentation | 10 days | Tue 11/16/21 | Mon 11/29/21 | 16 |
| 19 | | Updated Project Plan | 2 days | Tue 1/4/22 | Wed 1/5/22 | |

| | | | | | | | | | |
|---|---|---|---|---|---|---|---|---|---|
| 20 | | | Status Update Memo | 4 days | Thu 1/6/22 | Tue 1/11/22 | 19 | | |
| 21 | | | **Status Update Memo II** | **7 days** | **Wed 1/12/22** | **Thu 1/20/22** | **20** | | |
| 23 | | | Functional Prototype Demo/ Test Plan Presentation | 6 days | Wed 1/19/22 | Wed 1/26/22 | 32,33,34,35 | | |
| 24 | | | Team Health Assesment/Peer Evaluation | 3 days | Tue 2/1/22 | Thu 2/3/22 | | | |
| 25 | | | Ethics Reflection | 3 days | Tue 2/1/22 | Thu 2/3/22 | | | |
| 26 | | | **Status Update Memo III** | **10 days** | **Wed 1/26/22** | **Tue 2/8/22** | **23,21** | | |
| 27 | | | Human Interface Testing | 8 days | Wed 1/26/22 | Fri 2/4/22 | | | |
| 28 | | | Statuse Update Memo IV | 5 days | Wed 2/9/22 | Tue 2/15/22 | 26 | | |
| 29 | | | Status Update Memo V | 5 days | Wed 2/16/22 | Tue 2/22/22 | 28 | | |

| | | | | | | | | |
|---|---|---|---|---|---|---|---|---|
| 26 | | | **Status Update Memo III** | **10 days** | **Wed 1/26/22** | **Tue 2/8/22** | **23,21** | |
| 27 | | | Human Interface Testing | 8 days | Wed 1/26/22 | Fri 2/4/22 | | |
| 28 | | | Statuse Update Memo IV | 5 days | Wed 2/9/22 | Tue 2/15/22 | 26 | |
| 29 | | | Status Update Memo V | 5 days | Wed 2/16/22 | Tue 2/22/22 | 28 | |
| 30 | | | Status Update Memo VI | 5 days | Wed 2/23/22 | Tue 3/1/22 | 29 | |
| 31 | | | Demo Day/Design Review Presentation | 5 days | Wed 3/2/22 | Tue 3/8/22 | 30 | |
| 32 | | | COMSOL Simulation Testing | 6 days | Fri 1/7/22 | Fri 1/14/22 | | |
| 33 | | | Hardware Assembly | 6 days | Tue 1/4/22 | Tue 1/11/22 | | |
| 34 | | | Arduino Programming | 8 days | Tue 1/4/22 | Thu 1/13/22 | | |
| 35 | | | Breadboard Testing | 11 days | Tue 1/4/22 | Tue 1/18/22 | | |

*Figure 3: Above is a legend specifying each of the boxes on the Network Diagram.*

## Indications For Use

Based on the customer requirements and engineering specifications, we have developed the following Indications for Use to underline what the device should be specifically used for in your lab.

Our product is a syringe pump that delivers fluid at a controlled flow rate to cells in a microfluidic device. Its intended use is to aid in research that uses microfluidic devices with extracellular matrix gels containing human cells to test how the cells react to the fluid that is entering and exiting at a rate that mimics true physiology. The device is intended strictly for pushing fluids into microfluidic devices, not pulling fluids out. This research will be used greatly in the pharmaceutical field. The intended users of our product are trained researchers in the Microphysiological Systems Lab at Cal Poly SLO, and it is intended to be used inside of that lab only. It is to be used to put the cells under physiological shear stress as determined by data depending on experiment, and should be able to run for up to 9 days depending on the flow rate inputted by the user. Additionally, it is designed to be used with a 1 mL syringe.

*Note: We updated the IFU to specify that the pump can run for several days at a time only if the flow rate is substantially low, otherwise the linear actuator will run out of room to displace the syringe forward. Additionally, we clarified that it was designed for a 1 mL syringe, however it can be adapted for different sizes by printing alternative pusher blocks.*

## Customer Requirements

Following research on both intellectual property and academic resources that will help us in our design of the syringe pump, we discussed with you what exactly you want the syringe pump to do. We created a House of Quality in order to analyze what requirements you as the customer are requesting and how important each one is in comparison with the others. In this House of Quality we also compared the syringe pump we plan on making against two competitive models on the market.

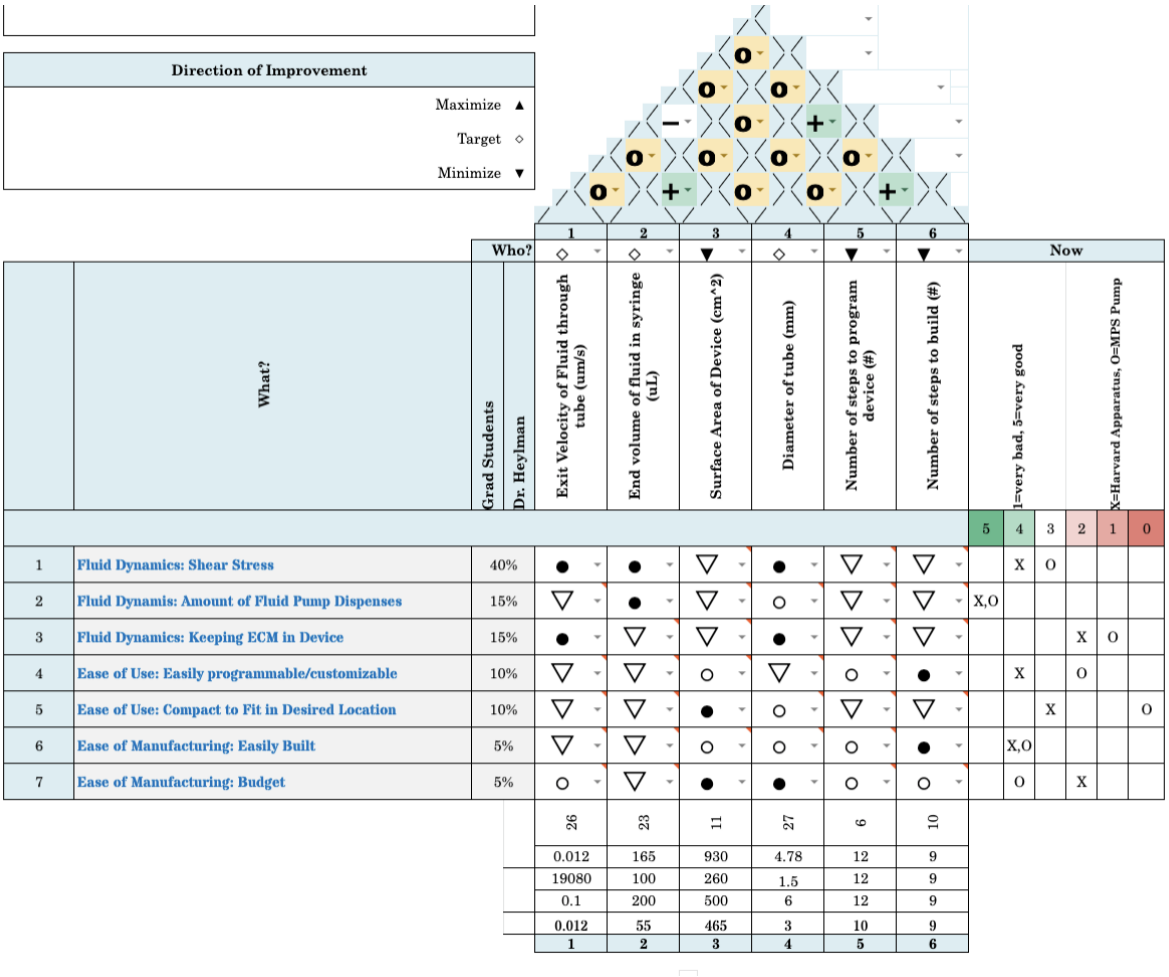



*Figure 4: An image of the House of Quality.*

**Table 3: Quality Function Deployment Summary**

| QFD Step | | |
|---|---|---|
| 1. | Identify Customers | MPS Lab grad students, Professor Heylman |

| | | |
|---|---|---|
| 2. | Determine the Customers Requirements | Specific shear stress on cells, controlled amount of fluid pump dispenses, keeping ECM in device, Easily programmable and customizable, compact, easily built, fairly low budget |
| 3. | Determine Relative Importance of Requirements | In same order, importance out of 100% broke down to 45%, 15%,15%, 10%, 10%, 5%, 5% |
| 4. | Identify and Evaluate the Competition | Existing design essentially meets customer requirements, looking to mimic and improve upon that |
| 5. | Generate Engineering Specifications | Controlled measured exit velocity of fluid, controlled end volume in syringe, surface area of device, diameter of tube,number of steps to program device, number of steps to build device |
| 6. | Relate Customer Requirements to Engineering Specifications | See the House of Quality in Customer Requirements |
| 7. | Set Engineering Specification Targets and Importance | See bottom of the House of Quality in Customer Requirements |
| 8. | Identify Relationships Between Engineering Specifications | Specific shear stress related to v, flow rate related to v and d, number of steps to program and number of steps to build positively correlated |

We collected this list of requirements and weighted them based on what we discussed was most important to you.

**Table 4: Customer Requirements**

| |
|---|
| Fluid Dynamics: Shear Stress that Mimics True Physiology |
| Fluid Dynamics: Accurate Flow Rate Delivered to Device |
| Fluid Dynamics: Keeping ECM in Device |
| Ease of Use: Easily programmable/customizable |
| Ease of Use: Compact to Fit in Desired Location |

## Specification Development

After receiving customer requirements from you as our sponsor, we have converted those requirements into measurable metrics that we will design the syringe pump to hit. For both flow rate and shear stress, our pump will be able to deliver a wide range of each. However, we are using the target flow rate and shear stress that you specified will be used most often for your research purposes.

**Table 5: Summary of Specification Development**

| Spec # | Parameter Description | Target | Tolerance | Risk | Compliance |
|---|---|---|---|---|---|
| 1 | Flow Rate | 0.76, 7.6, 75.8, 758, 1920, 9600, 28800 µL/hr | ±15% | M | T |
| 2 | Shear Stress | 3.60, 35.99, 358.91 µm/s | ±20% | M | T, A |
| 3 | Surface Area | 695 cm^2 | MAX | L | T, I |
| 4 | # Steps to Program | 5 steps | ± 1 | L | T, I |

*Note: We updated the tolerance for both the flow rate and shear stress testing after taking into account the difficulty in accurately testing these values and the variability that*

*occurs. Overall, we are still confident our device meets an even tighter tolerance but have expanded it due to complicated testing procedures.*

## Conjoint Analysis

Once we established our Indications for Use, we examined different features we would want to consider for our device (Table 6). We created conjoint cards with the three different features paired up in all of the different matchings. We had our classmates vote on which matches they think would be best in order to gather data. One everyone participated in the survey, we took the data and input it into excel (Figure 5). From there, we were able to run a multiple regression model to find out which of the variables were significant (Figure 6).

**Table 6: Levels and Factors for Conjoint Analysis**

| Factors | Level 1 | Level 2 |
|---|---|---|
| Display | Digital | Analog |
| Power Source | Electric | Battery-Powered |
| Surface Area | 1 ft^2 | 0.5 ft^2 |

Conjoint cards:
Analog, Battery, SA=1 ft^2
Analog, Electric, SA=0.5 ft^2
Digital, Battery, SA=0.5 ft^2
Digital, Electric, SA=1 ft^2

| What is your name? | What is your Group #? | What is your Group | [Analog, Battery, SA=1 ft^2] | [Analog, Electric, SA=0.5 ft^2] | [Digital, Battery, SA=0.5 ft^2] | [Digital, Electric, SA=1 ft^2] |
|---|---|---|---|---|---|---|
| Karoline Wucherer | 1 | Bone and Cartilage ( | 1 | 2 | 3 | 4 |
| Abby Jens | 7 | Live Imaging Cell Co | 2 | 4 | 3 | 1 |
| Elsa Bean | 7 | Live Cell Imaging Ga | 4 | 3 | 2 | 1 |
| Sophia James | 5 | Heart Pump Impelle | 4 | 1 | 3 | 2 |
| Ben Parmentier | 1 | Bone and Cartilage ( | 4 | 3 | 1 | 2 |
| Thomasina Hinkle | 1 | Bone and Cartilage ( | 4 | 3 | 2 | 1 |
| Amit Sharir | 5 | Impeller R&D | 4 | 2 | 3 | 1 |
| Kendall Gentzen | 5 | Heart Pump Impelle | 3 | 1 | 2 | 4 |
| Kerri Byrne | 6 | Live Cell Imaging He | 2 | 1 | 3 | 4 |
| Heidi Silk | 4 | FeverDots | 2 | 4 | 1 | 3 |
| Melanie Mitton | 3 | Semi-Automated M | 1 | 3 | 2 | 4 |
| Andrea ng | 3 | | 4 | 1 | 2 | 3 |
| Rita Grigorian | 4 | Fever Dots | 4 | 3 | 1 | 2 |
| Bryce Sakata | 3 | Semi-automated me | 4 | 2 | 1 | 3 |
| Brady Berg | 7 | Live Cell Imaging: Ga | 4 | 1 | 3 | 2 |
| Anna Frauenheim | 6 | Live Cell Imaging He | 4 | 2 | 3 | 1 |
| Simon Park | 6 | Live Cell Imaging He | 1 | 3 | 2 | 4 |
| Kameryn Laureano | 4 | FeverDots | 4 | 3 | 2 | 1 |

*Figure 5: A screenshot of the data gathered from peers and their rankings of each conjoint cards*

| SUMMARY OUTPUT | | | | | | | | |
|---|---|---|---|---|---|---|---|---|
| | | | | | | | | |
| *Regression Statistics* | | | | | | | | |
| Multiple R | 0.323941772 | | | | | | | |
| R Square | 0.104938272 | | | | | | | |
| Adjusted R Square | 0.065450254 | | | | | | | |
| Standard Error | 1.088411927 | | | | | | | |
| Observations | 72 | | | | | | | |
| | | | | | | | | |
| ANOVA | | | | | | | | |
| | *df* | *SS* | *MS* | *F* | *Significance F* | | | |
| Regression | 3 | 9.444444444 | 3.148148 | 2.657471264 | 0.055210538 | | | |
| Residual | 68 | 80.55555556 | 1.184641 | | | | | |
| Total | 71 | 90 | | | | | | |
| | | | | | | | | |
| | *Coefficients* | *Standard Error* | *t Stat* | *P-value* | *Lower 95%* | *Upper 95%* | *Lower 95.0%* | *Upper 95.0%* |
| Intercept | 4.333333333 | 0.678744088 | 6.38434 | 1.78967E-08 | 2.978920594 | 5.687746073 | 2.978920594 | 5.687746073 |
| X Variable 1 | -0.444444444 | 0.256541151 | -1.73245 | 0.08772729 | -0.956364342 | 0.067475453 | -0.95636434 | 0.067475453 |
| X Variable 2 | -0.277777778 | 0.256541151 | -1.08278 | 0.282730458 | -0.789697675 | 0.23414212 | -0.78969768 | 0.23414212 |
| X Variable 3 | -0.5 | 0.256541151 | -1.94901 | 0.055421493 | -1.011919897 | 0.011919897 | -1.0119199 | 0.011919897 |

*Figure 6: The excel output from the multiple regression model*

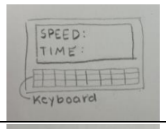
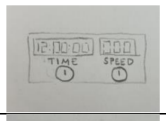
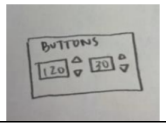
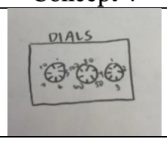
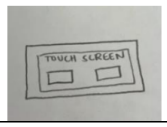
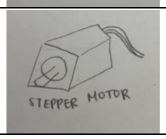
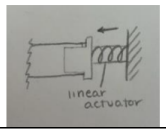
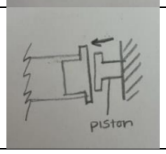
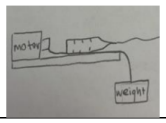
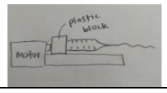
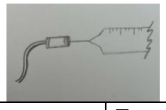
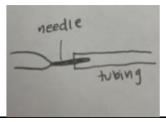
$Y=4.33-0.444x_{1}-0.278x_{2}-0.5x_{3}$

As seen in our results, all of our variables are negative, so a higher x leads to a lower y, meaning a lower numerical ranking. Additionally, because they are all negative, we know Level 2 is preferred in each factor. We are looking for the lowest numerical rankings, because 1 was assigned to the best combination of features, whereas 4 was assigned to the worst. Variable 3, which represents surface area, is the biggest in the negative direction, so that variable contributes most to a high rank customer attractiveness. Variable 1, which represents display, contributes to approximately 36.3% of customer attractiveness. Variable 2, which represents power source, contributes to approximately 22.7% of customer attractiveness. Lastly, Variable 3 contributes to approximately 40.9% of customer attractiveness. We are noting that none of the variables are significant because they all have a p-value of greater than 0.05. For Display Factor, digital is preferred. For Power Source, battery powered is preferred. Lastly, the surface area, 0.5 ft^2 is preferred.

## Morphology

After receiving feedback on the different features we were considering for our design, we created a morphology chart (Table 7). In this chart we considered all of the functions of our syringe pump. From there, we each brainstormed features that would address each of those functions. We illustrated each concept and added it under its respective function. We used this morphology chart to organize our ideas. From there, we used these ideas to create three ideas for concepts: Concept 1 (Figure 7), Concept 2 (Figure 8), and Concept 3 (Figure 9) seen below.

**Table 7: Morphology Displaying Concept Ideas for Syringe Pump Functions**

| Morphology | | | | | | |
|---|---|---|---|---|---|---|
| Product: Syringe Pump | | Organization Name : MPS Lab | | | | |
| Function | Concept 1 | Concept 2 | Concept 3 | Concept 4 | Concept 5 | Concept 6 |
| Inputting parameters on interface |  |  |  |  |  | |
| Power Source for Pushing |  |  | | | | |
| Mechanism for Pushing |  |  |  |  | | |
| Delivering Fluid into Tube |  |  | | | | |
| Team member: Lauren | Team member: Sophie | | Prepared by: Syringe Pump Team | | | |
| Team member: Nick | | | Checked by: | | Approved by: | |
| *The Mechanical Design Process*<br>Copyright 2008, McGraw Hill | | | | Designed by Professor David G. Ullman<br>Form # 15.0 | | |

Concept 1

The first element in this concept sketch is inputting the parameters through dials. Dials are commonly used and allow for easy interpretation and execution of the parameters. The second element is the power source for pushing, which is our stepper motor. A stepper motor powered by an Arduino will turn a certain amount depending on the parameters and cause the shaft to drive forward. The shaft will drive forward onto a pusher block, which is the mechanism for pushing. This will drive the fluid into the tube and through the needle. The needle will be inserted directly into the tube.

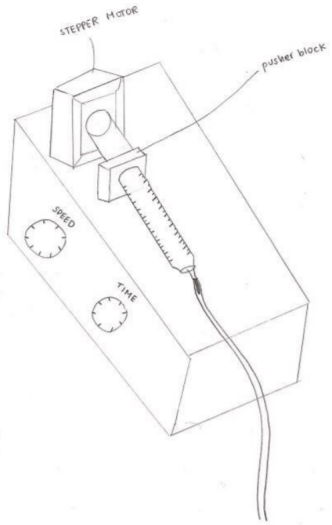



*Figure 7: A sketch of concept 1*

Concept 2

Concept 2 includes a linear actuator powering the syringe pump, pushing a block that makes contact with the syringe. This block pushes on the back of the syringe to release the contents in a controlled manner. The syringe is connected to a cylinder that has adjustable diameters so that the needle enters on one side, and the delivering tube enters in the other. This concept includes a screen with a keyboard where the user can insert parameters like exit velocity of the contents as well as time duration of the experiment. The keyboard includes numbers as well as arrows and select keys in order to navigate on the screen. The design includes a base that holds the linear actuator with the syringe and includes syringe stabilizers attached to the base of the syringe as well as the narrowed head of the syringe.

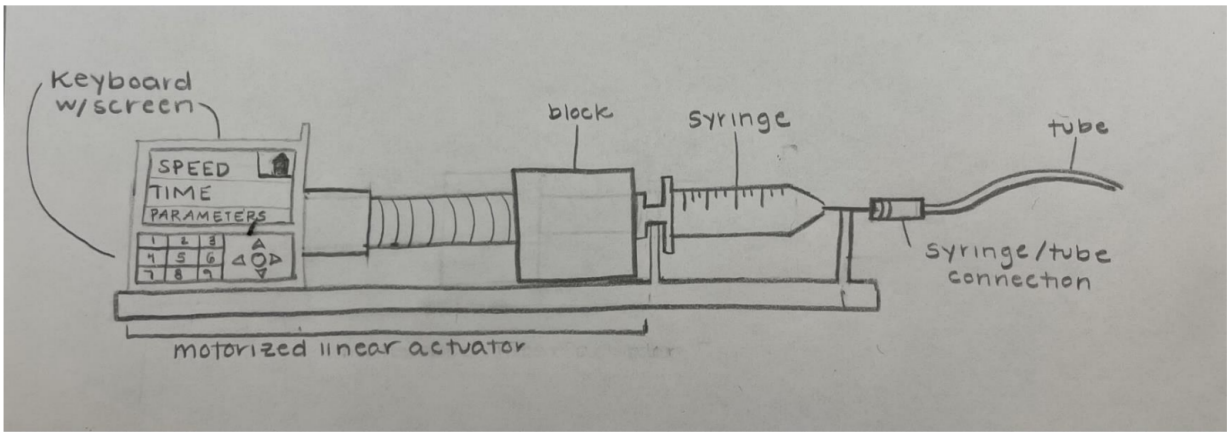



*Figure 8: A sketch of concept 2.*

Concept 3

Concept 3 includes a design that has a stepper motor that connects to a linear piston via a rack and gear system. Input parameters would be input via a touchscreen on the chassis of the device. To limit flow rate loss through the syringe, Concept 3 uses a needleless design, so a single diameter tube would be embedded into the syringe body. The touch screen would probably be an embedded android phone that communicates with the stepper motor via an Arduino board.

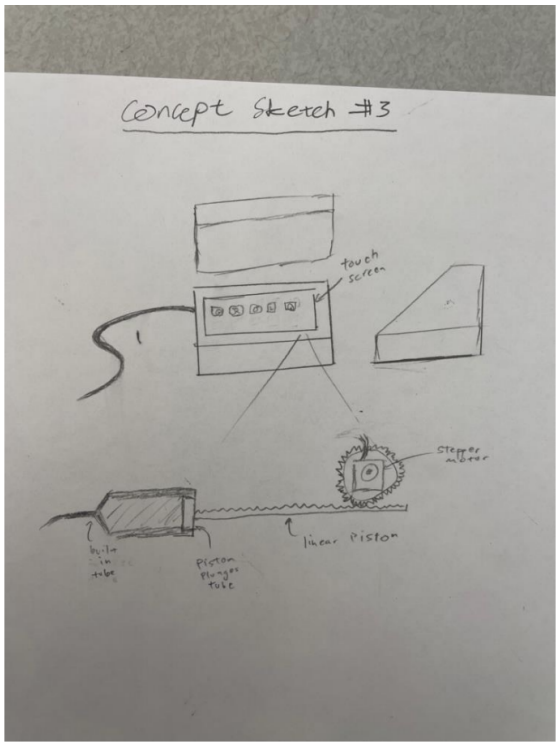



*Figure 9: A sketch of concept 3.*

**Concept Evaluation**

Once we had developed three concept designs for our syringe pumps, we used Pugh charts in order to compare the three concept designs against each other. Each member did three Pugh Matrices. The first Pugh Matrix compares Concept 1 against Concepts 2 and 3, the second Pugh Matrix compares Concept 2 against Concepts 1 and 3, and the last compares Concept 3 against concept 1 and 2. Then, we combined each of our first, second and third Pugh Matrices into three average matrices (Figure 10-12 seen below).

*Concept 1 Pugh Matrix Review*

According to this Pugh Chart analysis, concept 3 is the best, followed by concept 2, followed by concept 1. For controlling shear stress, our average marked concept 2 as worse because the linear actuator would be pre-programmed so it would not allow us to

have as specific velocity values and thus less exact shear stress imparted on the cells. For controlling the amount of fluid dispensed by the pump, concept 2 and 3 are better because the amount of turns for the worm gear to make the pusher block in concept 1 move forward are harder to compute. For keeping the ECM in the device, all three concepts are equally competent because the velocity is customizable in all three. For easy programmability, the touch screen in concept 3 makes it hard to program, but the keyboard in concept 2 might be easier to program and customize to exact values than turning the dials on concept 1. For the criteria of compactness to fit in the desired location, we thought all three could still meet this standard equally. For easily built, concept 2 would be easiest since the linear actuator is pre-programmed and we could just insert it. Concept 3 would be difficult to integrate a touch screen system to the pump. Lastly, concept 1 is best for budget because the dial system is simplest and a stepper motor is cheaper than a pre-programmed linear actuator.

## Pugh Matrix

| Pugh Matrix | | | | |
|---|---|---|---|---|
| **Critical Quality** | ***Weight*** | **Concept 1** | **Concept 2** | **Concept 3** |
| **Controlling shear stress** | 40 | 0 | -1 | 0 |
| **Controlling amount of fluid dispensed by pump** | 15 | 0 | 1 | 1 |
| **Keeping ECM in Device** | 15 | 0 | 0 | 0 |
| **Easily programmable/ customizable** | 10 | 0 | -1 | 1 |
| **Compact to fit in desired location** | 10 | 0 | 0 | 0 |
| **Easily Built** | 5 | 0 | 1 | -1 |
| **Low Budget** | 5 | 0 | -1 | -1 |

| Summary Table | | | |
|---|---|---|---|
| ***Total*** | 0 | -1 | 0 |
| ***Weighted Total*** | 0 | -35 | 15 |

*Figure 10: Average of our 3 Pugh Matrix Analysis for Concept 1*

*Concept 2 Pugh Matrix Review*

According to the Pugh Matrix comparing Concept 2 to the other two concepts, it showed that Concept 2 and Concept 3 both ended up with a net total of zero, while Concept 1 ended up with a net total of negative forty-five. While Concept 1 got a plus one for easily customizable and low budget, it got minus one for controlling shear stress (since the dials are less accurate compared to the other designs), for controlling amount of fluid released by the pump because it would be hard to calculate the number of turns

needed by the gear to make the pump run, and for being easily built because it does not come pre-programmed and would require the coding of an Arduino. Concept 3, while it received minus one for being difficult to build and not low budget, it got plus one for being easily programmable, ending with a net total of zero.

**Pugh Matrix**

| Pugh Matrix | | | | |
|---|---|---|---|---|
| **Critical Quality** | ***Weight*** | **Concept 2** | **Concept 1** | **Concept 3** |
| **Controlling shear stress** | 40 | 0 | -1 | 0 |
| **Controlling amount of fluid dispensed by pump** | 15 | 0 | -1 | 0 |
| **Keeping ECM in Device** | 15 | 0 | 0 | 0 |
| **Easily programmable/ customizable** | 10 | 0 | 1 | 1 |
| **Compact to fit in desired location** | 10 | 0 | 0 | 0 |
| **Easily Built** | 5 | 0 | -1 | -1 |
| **Low Budget** | 5 | 0 | 1 | -1 |

| Summary Table | | | |
|---|---|---|---|
| ***Total*** | 0 | -1 | -1 |
| ***Weighted Total*** | 0 | -45 | 0 |

*Figure 11: Average of our 3 Pugh Matrix Analysis for Concept 2*

*Concept 3 Pugh Matrix Review*

According to the Pugh analysis, Concept 2 was the best concept. Concept 2 obtained a weighted score of 35, while concept 1 obtained a weighted score of -60. Concept 2 may provide easily controllable shear stress rates due to its combined motor and linear actuator system, meaning that there is less error where systems interact with each other. Additionally, the motorized linear actuator provides good control of fluid out

of the system. Both concept 1 and 2 lost points in compact design, as the motor is in line with the actuator, increasing the design's size. These concepts both beat out concept 3 in being easily built, as concept 3 has a touch screen that must be connected to a more complex computing system in order to work properly.

**Pugh Matrix**

| Pugh Matrix | | | | |
|---|---|---|---|---|
| **Critical Quality** | ***Weight*** | **Concept 3** | **Concept 2** | **Concept 1** |
| **Controlling shear stress** | 40 | 0 | 1 | -1 |
| **Controlling amount of fluid dispensed by pump** | 15 | 0 | 1 | -1 |
| **Keeping ECM in Device** | 15 | 0 | 0 | 0 |
| **Easily programmable/ customizable** | 10 | 0 | -1 | 0 |
| **Compact to fit in desired location** | 10 | 0 | -1 | -1 |
| **Easily Built** | 5 | 0 | 1 | 1 |
| **Low Budget** | 5 | 0 | 0 | 1 |

| Summary Table | | | |
|---|---|---|---|
| ***Total*** | 0 | 1 | -1 |
| ***Weighted Total*** | 0 | 35 | -60 |

*Figure 12: Average of our 3 Pugh Matrix Analysis for Concept 3.*

Each of our charts represent the average score that each of us gave for that particular analysis.

After reviewing the combined scores, we decided that concept 3 is our front runner. It combines controllable shear stress with ease of use and programmability, and its motor not being in series with the piston allows us to minimize the footprint of the device. In the end, the final scores were 15 points for concept 3, 0 points for concept 2, and -105 points for concept 1.

After our analysis came to the conclusion of our front runner, we had to decide our front runner was feasible for the scope of this senior project. Because we are the ones manufacturing the syringe pump, we thought that an iteration of our optimal device without the touch screen might be better. We decided that for inputting the parameters, we will have our optimal design include a keypad with a display screen instead, as this is more reasonable for our knowledge base to manufacture and afford. This was an iteration for inputting our parameters that came in close second in our analysis, so we figured iterating to this would not drastically affect the attractiveness of our device nor would it affect adherence of our device to customer specifications. Additionally, we iterated to the stepper motor connected to a linear actuator. Upon further research, we found that this would be a more accurate and effective pushing mechanism for our syringe pump.

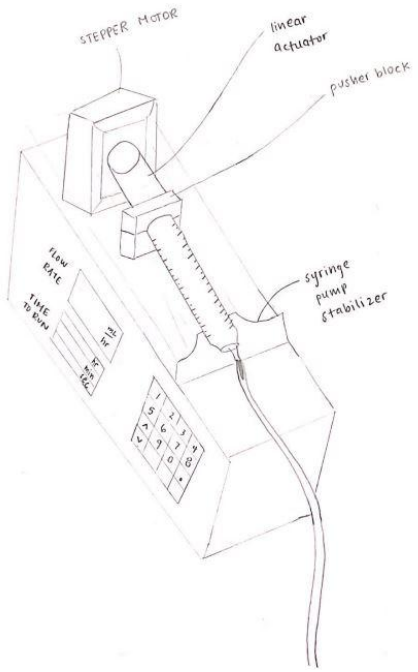



*Figure 13: A sketch of our final chosen design concept.*

**Conceptual Model**

Our conceptual model consists of a Comsol Multiphysics file of the microfluidic device in which the cell culture that our syringe pump feeds into lives. The model was originally created by the graduate students in the microfluidics lab. They created the model in hopes of growing cell cultures inside of the device. However, the extracellular matrix that houses the cell culture keeps getting pushed out during testing. To combat these, we needed data to ensure that our syringe pump has the appropriate volumetric

flow rate to entrance/exit size ratio. If the flowrate is too great and the entrance too small, the velocity will be too high and the ECM will come out. If the flowrate is too low and the entrance too large, it will not impart the physiological shear stress necessary for the cells to grow.

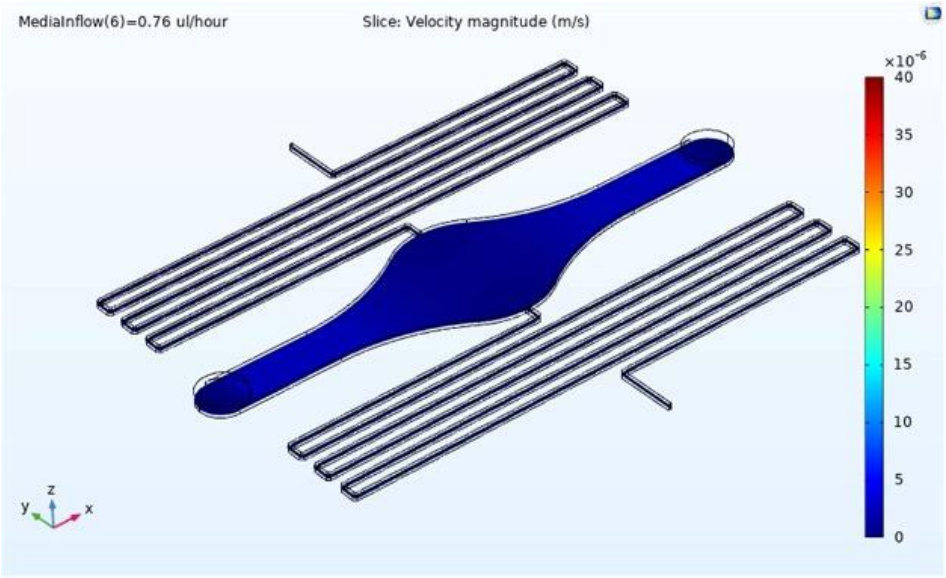



*Figure 14: Surface Plot of velocity through the microfluidic device performed on Comsol.*

In Comsol, three physics models were created to model loading of ECM into the device, perfusion of fluid through the device, and the porosity characteristics needed to simulate fluid through ECM. For the porosity of the ECM. When running the simulations for our specific results, our study used perfusion of the fluid through the device with the porosity characteristics of the ECM. This gave us results for interstitial fluid flow.

We established a boundary condition on the too large circles seen in the center of the device. One of those plugs was deemed the entrance and the other was the exit. Sabrina from the lab said that they currently run the pump at a volumetric flow rate of .76 uL/hr. We set the boundary condition of the device to the following: A laminar inlet flow of .76 uL/hr at one end of the device, and a matching laminar outlet flow at the other end. This flow rate pushes the ECM out, so we ran a parametric sweep with the volumetric flow rate starting at .5microliters/hr and jumping up by .05 until it reaches .76 uL/hr (.75 was omitted in favor of .76). Following the simulation, the results for fluid velocity as it relates to the volumetric flow rate can be seen in the figure below. Additionally, the fluid velocity was calculated by dividing the volumetric flow rate by the area of the syringe pump tube (17.87mm^2). The values are incredibly small, but interstitial fluid flow is a slow process.

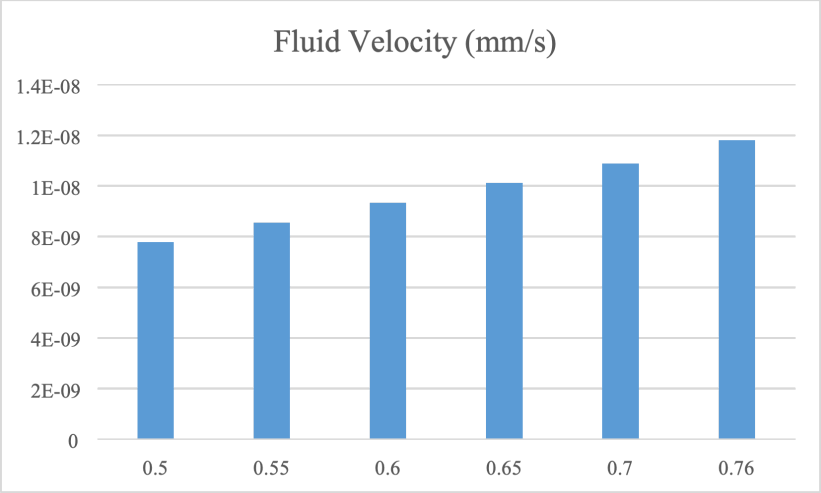



*Figure 15: Bar graph comparing Volumetric Flow Rate with Fluid Velocity*

Because velocity through the device is very slow and nearly uniform throughout the device, I used a surface average to evaluate the shear stress values within the device. A graph comparing shear stress to flow rate can be seen in the figure below.

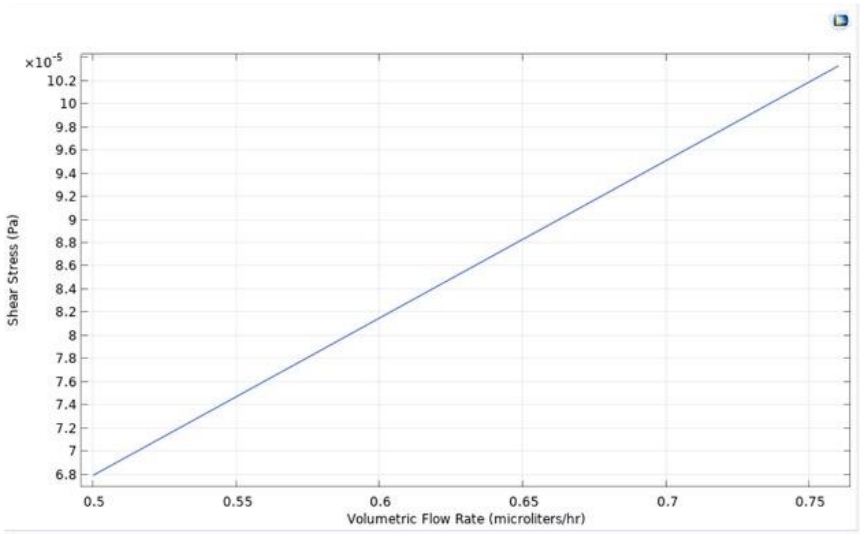



*Figure 16: A line graph displaying the positive correlation between shear stress and volumetric flow rate*

The wall shear stress value is already incredibly small at the highest volumetric flow rate, coming in at 10.2E-5 Pa. This value is not even remotely close to the physiological shear stress values that are expected. A study that studied interstitial fluid flow through a microfluidic device housing ECM and mesenchymal stem cells in an attempt to induce osteogenesis claims an expected shear stress of .0135 Pa (Kim). This expected shear stress value is 100x greater than our simulated value. However, their flow rate was far greater than ours. They used .274 μL/min compared to our .76μL/hour.

## Budget

We have created the budget outlining the different components for our design as well as the grand total we plan to spend on our prototype. Below is the budget we had originally planned and a summarized version of our actual final budget.

**Table 8: Planned Budget**

| Item Description | Product Number | Quantity | Cost/Unit | Total Cost |
|---|---|---|---|---|
| 4.78 mm Tubing | UPC-1650L | 5 | $5.00 | $25.00 |
| Food Coloring | B07HR2XWP6 | 1 | $5.00 | $10.00 |
| Pusher Block | Creating own | 1 | $5.00 | $5.00 |
| Syringe Holder | Creating own | 1 | $5.00 | $5.00 |
| Syringe Chassis | Creating own | 1 | $15.00 | $15.00 |
| Linear Guide Slide Table Ball Screw Motion Rail CNC Linear Guide Stage Actuator Motorized Nema 23 Stepper Motor | FSL40E15005C7 | 1 | $155.00 | $155.00 |
| Arduino Starter Kit | B01EWNUUUA | 1 | $63.00 | $63.00 |
| Stepper Motor Drive | B06Y5VPSFN | 1 | $29.00 | $29.00 |
| DC Power Supply | LED0060S360W24V15A | 1 | $30 | $30 |
| | | | **Total:** | $337.00 |

**Table 9: Actual Final Budget**

| Item Description | Quantity | Cost/ Unit | Total Cost |
|---|---|---|---|
| 4.78 mm Tubing | 3 | $ - | $ - |
| Pusher Block | 1 | $ - | $ - |
| Syringe Holder | 1 | $ - | $ - |
| Linear Guide Slide Table Ball Screw Motion Rail CNC Linear Guide Stage Actuator Motorized Nema 23 Stepper Motor | 1 | $ 155.00 | $ 155.00 |
| Arduino Starter Kit | 1 | $ 63.00 | $ 63.00 |
| Stepper Motor Driver | 1 | $ 29.00 | $ 29.00 |
| DC Power Supply | 1 | $ 30.00 | $ 30.00 |
| 1 x 1.5' Polycarbonate Sheet | 2 | $25.98 | $ 51.96 |
| Hinge Pin | 1 | $9.04 | $ 9.04 |
| Tygon Tubing | 1 | $ 66.50 | $ 66.50 |
| Dispensing Tubes | 1 | $ 12.50 | $ 12.50 |
| 4x Objective | 1 | $190 | $190 |
| | | **Total** | **$ 607.00** |

*Note: The budget was updated since the last report to include the items we have ordered as well as the items we have discovered we need, including test plan materials. Generally our test plan materials are items that we already have, so they do not add to our budget. We also updated this section to include our planned budget versus our actual final budget.*

**Detailed Design**

After reconsidering our front runner concept, we decided that the rack and gear system may not be the most efficient method for powering our syringe pump. We conducted further research and found a motorized linear actuator that we have decided to go with. We went with the FUYU FSL40 Linear Guide Slide Table Ball Screw Motion Rail CNC Linear Guide Stage Actuator Motorized Nema 23 Stepper Motor. This includes a pusher block that the stepper motor driver. From there, we needed to make an adaptor for the motorized linear actuator to hold the syringe. We used Solidworks to make two different pieces. The first one was to be attached to the top of the pusher block (see Figure 22). It has a slot where the back of the syringe will rest in; this is the part that needs to be pushed forward to cause the contents in the syringe to be distributed. As the motorized linear actuator is powered, the built in pusher block will move forward which will cause the attachment to also move forward so that the back of the syringe is getting the appropriate amount of applied pressure to release the syringe contents at the specified parameters. The second piece we built in Solidworks was an adaptor to the end of the motorized linear actuator to hold the syringe base (see Figure 19). This was needed in order to stabilize the syringe; it attaches to the back of the barrel of the syringe and is there to hold it in place while the pusher blocks to pressing out the contents. We also decided to make a large chassis on Solidworks that encompasses the entirety of the device (see Figure 23). This will protect the user from any pinch points while using the syringe and allow for greater mobility of the device. The chassis is a large box with a hinged lid that is easily opened and closed. We included an air vent to account for any heat that comes from the motor. It will be made out of polycarbonate so that the user can see the syringe pump in action and assure that everything is running smoothly and because polycarbonate is easily sanitized with IPA.

The other half of the design is the electrical side. This begins with an Arduino; we are using an Arduino UNO R3. From there, we have to consider how the user is to input parameters and how they will be able to see what they inputted. In our Arduino starter kit we received several tactile buttons that will be used to input the desired flow rate for the syringe pump. It also included an LCD screen, which is where the parameters will be displayed. These are connected to the Arduino with wires and are controlled by code written on the computer and downloaded onto the Arduino. The Arduino communicates with the stepper motor via a stepper motor driver. Our stepper motor driver will be the DM542T, a large driver that is capable of up to 25000 microsteps per revolution. This

driver allows us to achieve a high level of resolution, which is needed for the tiny flow rates that will be used in your lab. To connect the three parts we have to use wires to connect the Arduino to the driver, and then use wires to connect the driver to the stepper motor. The stepper motor is attached to the linear actuator which will push the pusher block and release the contents of the syringe.

*First Design Iteration*: Below are the first iterations of each piece of the project that we are manufacturing. First is the pusher block attached to the linear actuator via 4-M4X8 screws (Figure 17). Next is the first iteration of the syringe base holder which ended up being our final iteration (Figure 19). Lastly, our initial iteration of the chassis can be seen below as well. Our chassis design changed significantly as our material and thus manufacturing technique was altered throughout the design process (Figure 21).

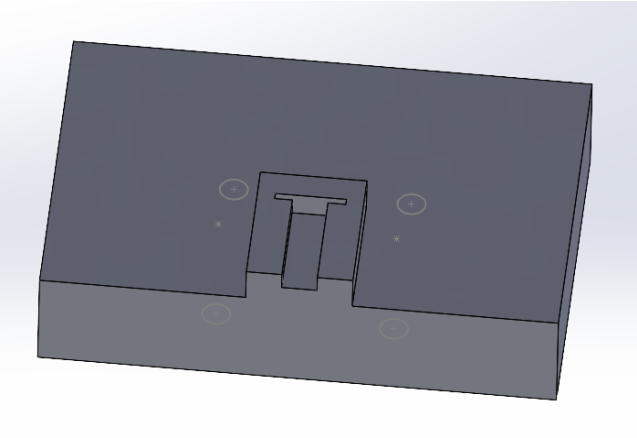


*Figure 17: SolidWorks model of pusher block adapter.*

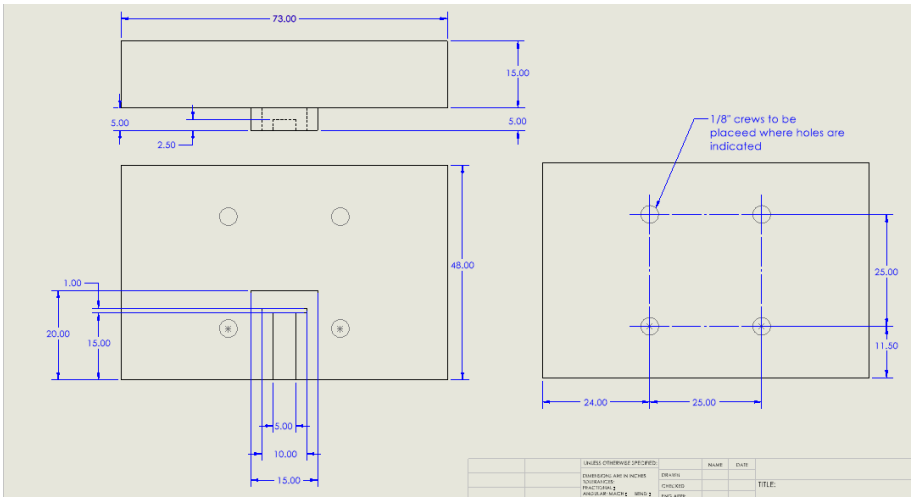



*Figure 18: Drawing of the pusher block adaptor created in Solidworks.*

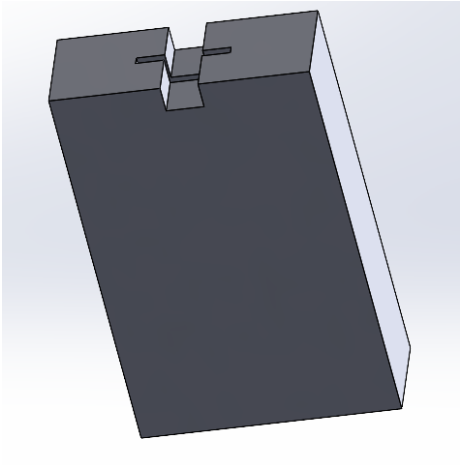


*Figure 19: SolidWorks model of base holder adapter.*

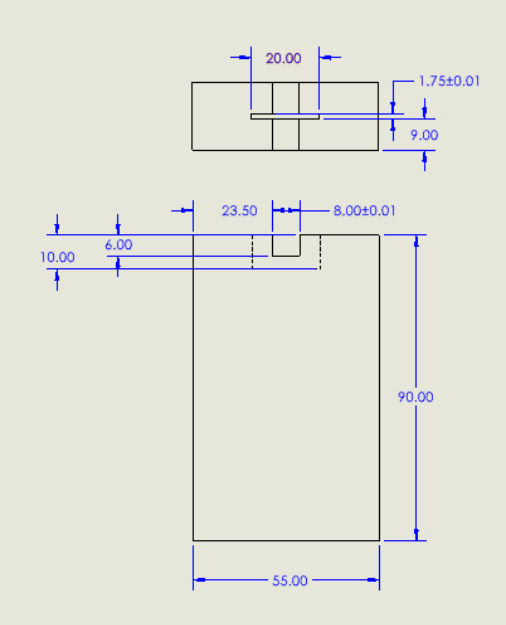



*Figure 20: An image of the adaptor for the back of the motorized linear actuator that will stabilize the syringe pump created in Solidworks.*

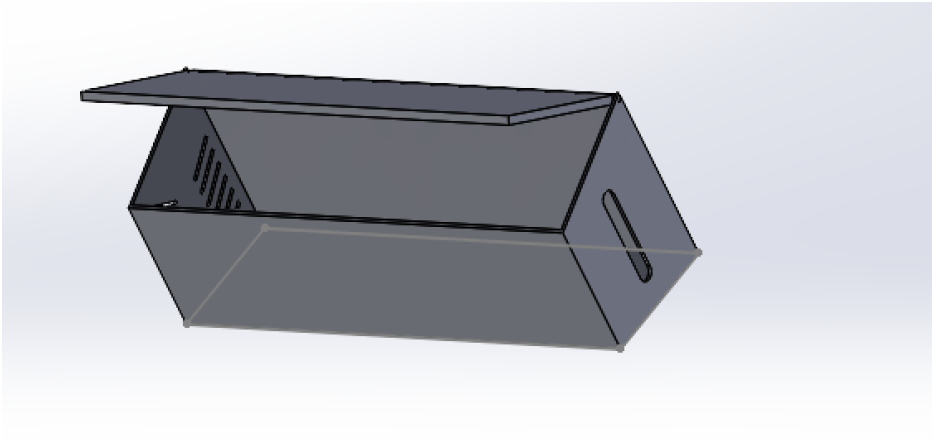


*Figure 21: Original SolidWorks model of the chassis.*

*Second Design Iteration:* Our second design iteration updated the size of the cuts where the syringe is inserted, as we did not allow enough tolerance in the first iteration, and thus it was difficult to insert the syringe. Additionally, we adjusted the thickness of the back so that the 12 mm screws would pierce through the block and be fully inserted into the existing holes (Figure 22). Additionally, after changing our manufacturing techniques, our updated chassis design can be found below (Figure 24). We made changes in order to make the manufacturing easier and more feasible in our timeline, as some of the features (i.e. tilted lid and vent shapes) added complications without adding any real value in terms of meeting our engineering specifications

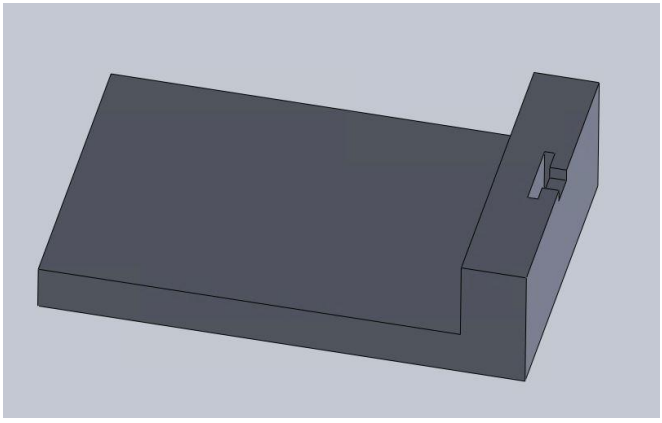


*Figure 22: Updated Pusher Block Solid Model*

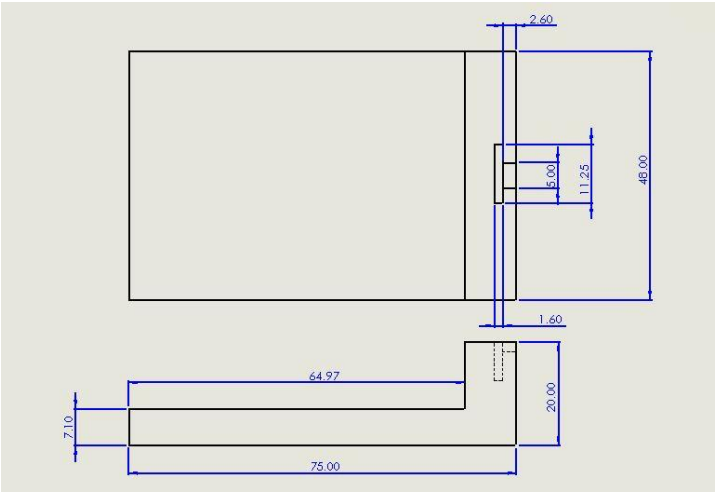



*Figure 23: Updated Pusher Block Solid Model Drawing*

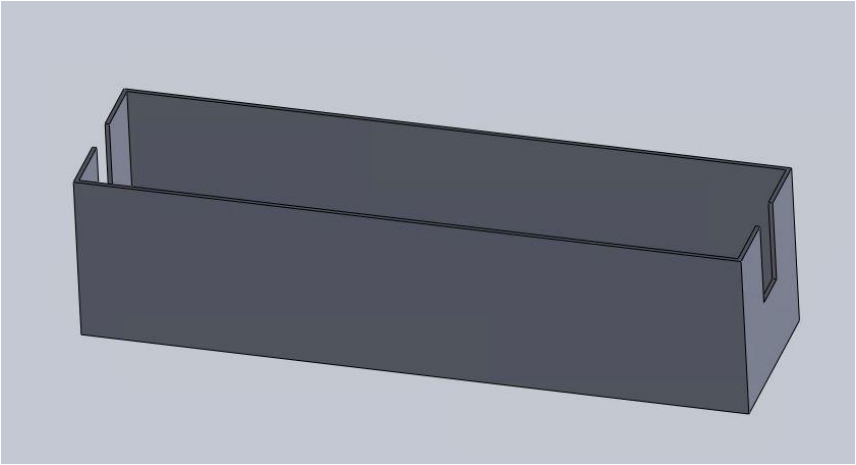


*Figure 24: Solid Model of the Final Chassis Design*

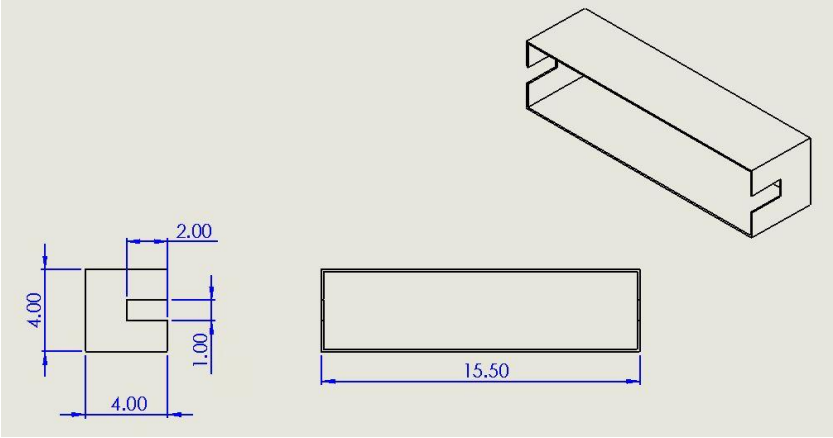



*Figure 25: Drawing of the Final Chassis Design*

Figure 26 below is the drawing of the assembly that displays how each element will fit together in our syringe pump.

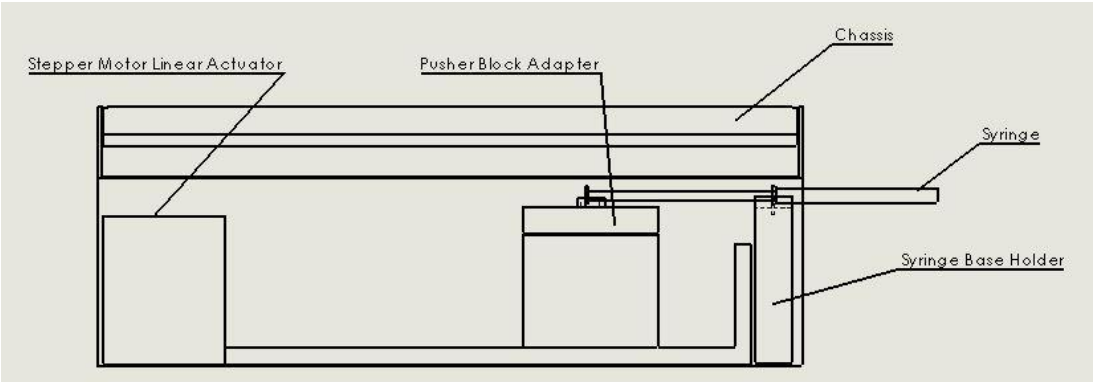



*Figure 26: An image of the chassis that will encase the linear actuator and all other internal components of our syringe pump.*

Below is the basic schematics for the electrical components of our design. Originally, we planned for a keypad  to take user inputs like fluid flow rate and time of the experiment. The two tactile buttons are connected to an Arduino UNO R3 and communicate with an LCD screen. The LCD screen uses a step down 10kΩ resistor connected to the screen's cathode; additionally, the two tactile buttons use 10kΩ resistors connected to their anode. This design can be summarized in Figure 26 below.

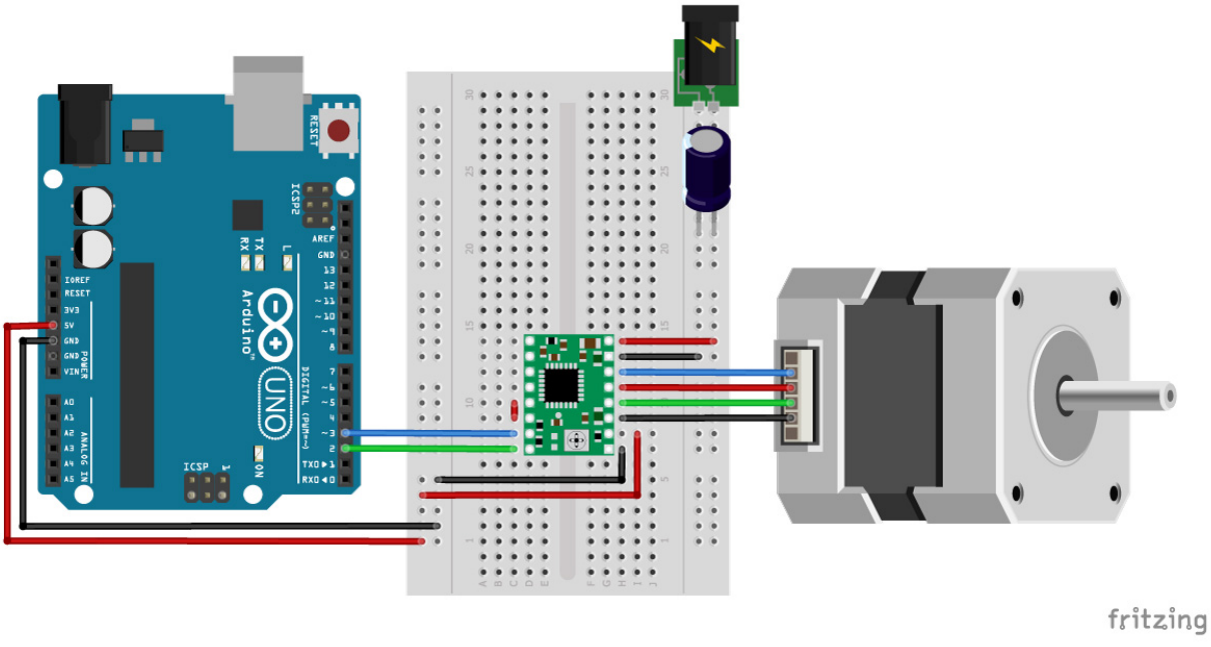



*Figure 27: The Arduino connected to the stepper motor driver which is connected directly to the stepper motor.*

In terms of the interfaces of our device, we planned to use a variety of techniques throughout our design planning phase. Our final plan came down to using Loctite super glue to connect the walls of the chassis, as it is impact and water resistant and was recommended specifically for polycarbonate. We decided to use velcro to adhere the electrical components to the lid of the chassis in order to secure them but allow them to be removed if needed to re-attach components or trouble shoot the device when malfunctioning. As stated above, 4-M4X8 screws were used to attach the pusher block adapter after having drilled holes in our 3D printed part. Loctite super glue was also used to attach the syringe base holder adapter to the end of the linear actuator.

*Note: The Detailed Design section was updated to include the updated design of our pusher blocks and chassis, as well as the electrical components. We also updated the way that the components interface with one another in our final prototype.*

**Prototype Manufacturing Plans**

The first part that we plan on manufacturing is the two adapters to the linear actuator. These will be 3D printed in Mustang 60, which has six functional JGAurora 5S printers. The following is the detailed manufacturing steps in creating the 3D printed parts and attaching them to the linear actuator.

1. Download the Cura application to your device.
2. Select JGAurora 5S as the printer and PLA as material.
3. Save SolidWorks parts as STL files.
4. Upload STL file to JGAura 5S via flash drive.
5. Ensure the machine is working properly by watching the first layer of PLA being applied.
6. Pick up parts when they are finished.
7. Measure placement of holes on pre-existing pusher block.
8. Drill holes onto 3D printed pusher block adapter in those locations.
9. Screw in the 4-M4X8 screws to secure attachment.
10. Secure syringe base holder adapter to linear actuator via epoxy.

The next part in manufacturing is the chassis. After deliberating, we decided on making the chassis out of polycarbonate rather than acrylic to due the brittle properties of acrylic when cleaned with IPA. We went to Home Depot and purchased a 2 x 1.5 foot sheet of polycarbonate that is enough to manufacture the entire chassis.

*First Iteration*: Because polycarbonate cannot be cut by the laser cutter, we decided to water jet the pieces in Mustang 60. We filled out the request form specifying what parts we wanted cut, how many of each, and when we needed it to be finished. We also

uploaded DXP files of each of the walls to specify the shape and dimensions of each wall of the chassis. Once all six of the walls are cut we will then use hot glue to assemble the chassis. We will attach the lid via hinges so that the user can easily access the syringe and change out when needed. The following is the detailed manufacturing steps in creating the chassis:

1. Obtain sheets of polycarbonate
2. Create DXP files of each of the types of walls needed to be cut
3. Fill out form for water jetting including DXP files and dimensioned drawings of each wall
4. Once the pieces are cut, glue the five sides together
5. Attached the hinges to the back of the chassis and the lid so that it can swing open and close

*Second Iteration:* Our first chassis fell apart when it was hot outside because the hot glue we used to adhere the pieces together melted. When we went back to re-glue the pieces they were covered in chunky glue clumps that were difficult to remove. Because of this, we decided to start over. We traced the pieces of polycarbonate ahead of time, this time accounting for a little extra tolerance to have the two sides sitting on top of the base piece. We used different power tools to cut the sheets at the Aero Hanger and then used super glue to attach the sides together. The following is the detailed manufacturing steps in creating the chassis for the final iteration:

1. Obtain a sheet of polycarbonate.
2. Trace out the six different sides: the two smaller side panels, the base panel, the lid, and the two edge pieces where the syringe sticks out.
3. Using a Table Saw cut the long pieces using the traces made in the previous step so that the long cuts have been made.
4. Then, using the Vertical Band Saw, cut the shorter cuts needed to finish the sides.
5. Next, use the Wire Saw to make the cut outs that allow for the syringe to poke out and to create air flow.
6. Then, use super glue and attach all of the panels together.
7. Lastly, attach the hinges to the lid and one of the sides so that the lid can open and close with ease.

The chassis will encase the motor, which will simply be placed inside the chassis so that it can be removed to reload the syringe if need be. The following electrical components will be velcroed to the back of the chassis, and the keypad and LCD screen will be velcroed on the front of the chassis so that parameters will be easily inputted and seen by the user.

On the electrical side of the device, there are various different components that must be connected in order to get the stepper motor to execute an accurate pushing motion based on the parameters inputted. These include connecting the arduino to the driver, connecting the keypad to the arduino and the LCD screen to the buttons, connecting the driver to the motor, and finally connecting the driver to the DC power supply. Below are the steps to manufacturing the electrical side of the device.

To begin, a 24V unregulated industrial power supply is used to power the DM542T Stepper Motor Driver connected to a Nema 23 2.0A stepper motor. An Arduino UNO R3 is used as the controller for the device. A common cathode is used, with the PUL+ and DIR+ pins connected to the Arduino's 5V pin. The PUL- pin (used to control the Pulses that induce movement of the motor) is connected to the 7 pin, and the DIR- (used to control the direction of the motor) is set to the 6 pin. In Arduino IDE, a simple sketch was created to induce constant, uniform movement of the linear slide. The direction of the device is chosen by setting the DIR- pin to an either "HIGH" or "LOW" setting. "HIGH" makes the motor push the block forward and "LOW" makes the motor pull the block backwards.

The speed at which the motor rotates is decided by one of two factors. Located on the DM542T Stepper Motor Driver, there are a set of DIP switches used to control the amperage to the motor and the steps per revolution. Since our motor is a 2.0A motor, the first 3 DIP switches are set to the OFF/OFF/ON positions. This ensures that the proper current flows to the motor and prevents the motor from being burnt out. The last 4 DIP switches control the microsteps per revolution; for the functional prototype, we elected to use 25000 microsteps per revolution, the slowest setting. This setting provides the greatest level of control, but uses the most power as a motor spinning more slowly requires a greater amount of torque. For lower resolution jobs, a setting with a lower amount of microsteps per revolution is possible by manipulating the DIP switches. An additional way of manipulating the motor speed is by increasing or decreasing the time between pulses to the motors coils. As seen in Figure 21, by changing the number within the delayMicroseconds function, the user can increase or decrease the pulses sent per second.

```
sketch_jan22b
void setup() {
  // put your setup code here, to run once:
pinMode(8,OUTPUT);
pinMode(9,OUTPUT);
digitalWrite(9,HIGH); // LOW to bring the slide back and HIGH to push the slide forward
}

void loop() {
  // put your main code here, to run repeatedly:
digitalWrite(8, LOW);
digitalWrite(8,HIGH);
delay(3);
}
```

*Figure 28 - The Arduino sketch used to power the motor, subject to heavy modification as the project progresses.*

Below are the simplified steps in connecting the electrical parts of our device.

1. Connect 24 V power supply to an outlet.
2. Connect the power supply to the DM542T Digital Stepper Motor Driver via screws, being sure to connect the green cable to ground, black to positive, and white to neutral.
3. Connect wires from the stepper motor to the Arduino UNO, with the PUL+ and DIR+ pins connected to the Arduino's 5V pin and PUL- pin (used to control the Pulses that induce movement of the motor) is connected to the 9 pin, and the DIR- (used to control the direction of the motor) is set to the 8 pin. Set the DIR- pin to "HIGH".
4. Connect wires from the aplastirduino to the Nema 23 2.0A stepper motor.
5. The LCD screen was connected to pins 2 through 7. To alter the intensity of the LCD screen, a 10k potentiometer was connected to the screen.
6. Buttons were connected to pins 12 and 13 to control the input flow rate. The buttons are connected to 10k Ohm resistors.

The arduino sketch code uses a specified delay between electrical pulses to the motor to create fluid flow. To convert desired flow rate to the necessary delay value, a simple conversion was completed which can be found in Equation 1 below.

$$Q = \left(\frac{1000}{x}\right) * \left(\frac{1\ revolution}{25{,}000\ pulses}\right) * \left(\frac{0.2\ mL}{1\ revolution}\right) * \left(\frac{1000\ uL}{1\ mL}\right) * \left(\frac{3600\ sec}{1\ hr}\right) \quad \textbf{(1)}$$

*x=delay*

*Q=Volumetric Flow Rate*
*Note: Since our last report we added steps for connecting the screen and buttons to the arduino. We also added details about the code uploaded to the arduino.*

## Hazard and Risk Identification

Below is the risk hazard assessment that our team put together when first conceptualizing our design, in an attempt to recognize potential dangers and add precautions into our design rather than discovering the hazards after we have already built it.

**Table 10: Hazard and Risk Identification**

| |
|---|
| Team: Syringe Pump Advisor: Dr. Heylman |
| 1.Will any part of the design create hazardous revolving, reciprocating, running, shearing, punching, pressing, squeezing, drawing, cutting, rolling, mixing or similar action, including pinch points and sheer points? **YES** |
| 2. Can any part of the design undergo high accelerations/decelerations? **NO** |
| 3. Will the system have any large moving masses or large forces? **NO** |
| 4. Will the system produce a projectile? **NO** |
| 5.Would it be possible for the system to fall under gravity creating injury? **YES** |
| 6.Will a user be exposed to overhanging weights as part of the design? **NO** |
| 7. Will the system have any sharp edges? **YES** |
| 8. Will any part of the electrical systems not be grounded? **NO** |
| 9. Will there be any large batteries or electrical voltage in the system above 40 V? **NO** |
| 10. Will there be any stored energy in the system such as batteries, flywheels, hanging weights or pressurized fluids? **NO** |
| 11. Will there be any explosive or flammable liquids, gasses, or dust fuel as part of the system? **NO** |
| 12. Will the user of the design be required to exert any abnormal effort or physical posture during the use of the design? **NO** |
| 13. Will there be any materials known to be hazardous to humans involved in either |

| |
|---|
| the design or the manufacturing of the design? **NO** |
| 14. Can the system generate high levels of noise? **NO** |
| 15. Will the device/system be exposed to extreme environmental conditions such as fog, humidity, cold, high temperatures, etc? **NO** |
| 16. Is it possible for the system to be used in an unsafe manner? **YES** |
| 17. Will there be any other potential hazards not listed above? If yes, please explain in reverse. **NO** |

**Table 11: Summary of Planned Corrective Safety Actions**

| Description of Hazard | Planned Corrective Action | Planned Date | Actual |
|---|---|---|---|
| This design includes a pusher block that is pushing against the back of the syringe, creating the potential for pinch points. | In order to prevent this hazard, we plan on creating a protective case that contains the whole device while it's running. It can be removed to reload the syringe, but it is only to be removed when the device is off. | November 17 | Designed the chassis in December, finished assembling the chassis on January 21, 2022. |
| This system is small and can be easily moved. It is designed to sit on the counter so it could potentially be knocked off the edge onto someone. | In order to prevent this hazard, we will include a written warning to keep the device away from the edge of the surface where it is resting. | February 20 | Added a written warning in the Operation's Manual on January 13, 2022. |

| | | | |
|---|---|---|---|
| This system involves a syringe, and therefore includes sharp edges, opening for the potential injury. | In order to prevent this hazard, we plan on creating a protective case that contains the whole device. This will prevent someone from accidentally touching the sharp edge of the needle. There will also be tubing attached to the sharp needle at all times. | November 17 | Designed the chassis in December, finished assembling the chassis on January 21, 2022. |
| This system is intended for lab use only. Specifically, it is not intended for in vivo human use. | In order to prevent this hazard, we will include a written warning that this device is not to be used for in vivo human use. | February 20 | Added a specification about this in the Indications for Use in November, and a note in the Operation's Manual on January 20. |
| Our pump uses a motor and a DC Power Supply that may cause heat generation. | We plan to design air vents into the chassis and purchase a Power Supply that has a fan built within it, which will lie outside of the chassis for max airflow. | January 13 | Water jetted a big square into the polycarbonate wall for ventilation on January 13 and purchased a Power Supply with a self regulating fan in December. |

Overall, we feel that we have effectively mitigated the risk of using our syringe pump, and feel confident that we can test it safely and that you as our end user will remain safe while implementing the product into your research.

**Operation Manual / Instructions for Use**

The operation of the syringe pump should be fairly simple for any user in the MPS lab. The chassis pictured below is there to keep the user safe from pinch points while the syringe pump is operating.

When the pump is not operating, open the lid of the chassis to remove the syringe from the adapters and fill with desired fluid.

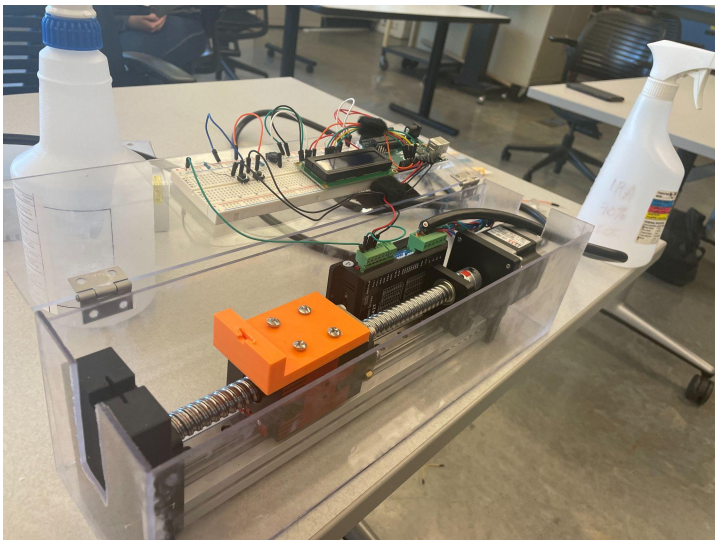


*Figure 29: Image of the chassis open when loading syringes and inputting parameters.*

Secure the loaded syringe back into the holes of the adapters and close the lid of the chassis. Make sure the syringe is placed securely and aligned horizontally with the tubing.

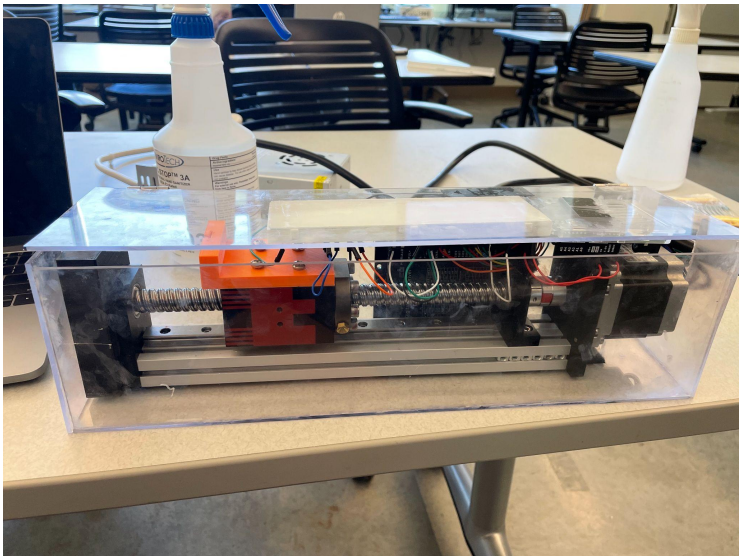


*Figure 30: Image of the chassis closed once syringe is loaded*

Be sure to keep the tubing on the needle of the syringe while loading medium to avoid exposing the sharp edge. Insert desired flow rate via the buttons, noting the units that are being delivered. The LCD screen will display what parameters you are entering as you enter them.

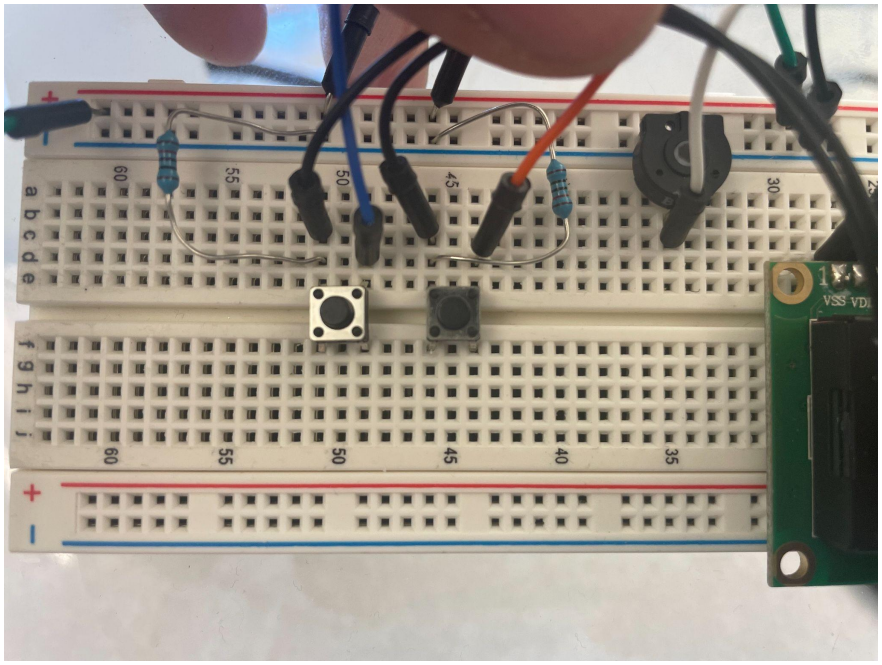


*Figure 31: Image of the buttons used to input parameters.*

Once you have inserted parameters, turn on the power supply and briefly watch the movement of the pump and the syringe to ensure parts are securely fastened and

working properly. Be sure to avoid placing the syringe on a surface that is near an edge, ensuring that the syringe will not be unintentionally displaced and cannot fall on the user. Additionally, keep the electrical components of the syringe away from common activity to ensure no dispacing of the wires or harm done to the user.

The big vent in the chassis behind the motor was designed to mitigate any heat generated during the process; however, the small flow rates desired by the MPS Lab is not expected to cause heat generation that would be of any harm to the user.

## Test Plans

We have created a separate Network Diagram as well as a summary table specifically for our test plans to organize our tasks and understand the critical steps and materials needed. The critical resources that we need include microbeads, a microscopic system with a camera that measures small velocities, and a sensitive scale. We have confirmed with Professor Heylman that the lab has microbeads, and we have confirmed with the grad students who work in the MPS Lab that they have a scale that measures milligrams, which is the sensitivity we need for running the flow rate testing. Dr. Hawkins agreed to train us in using the LabSmith Micro Particle Image Velocimetry microscope in his lab, which provides the x- and y-velocities of the microbeads being pushed through the microfluidic device via our pump. We have organized the testing chronologically from most elaborate test to least elaborate tests, but essentially no one test depends on or relies on another. They all can be performed independently.

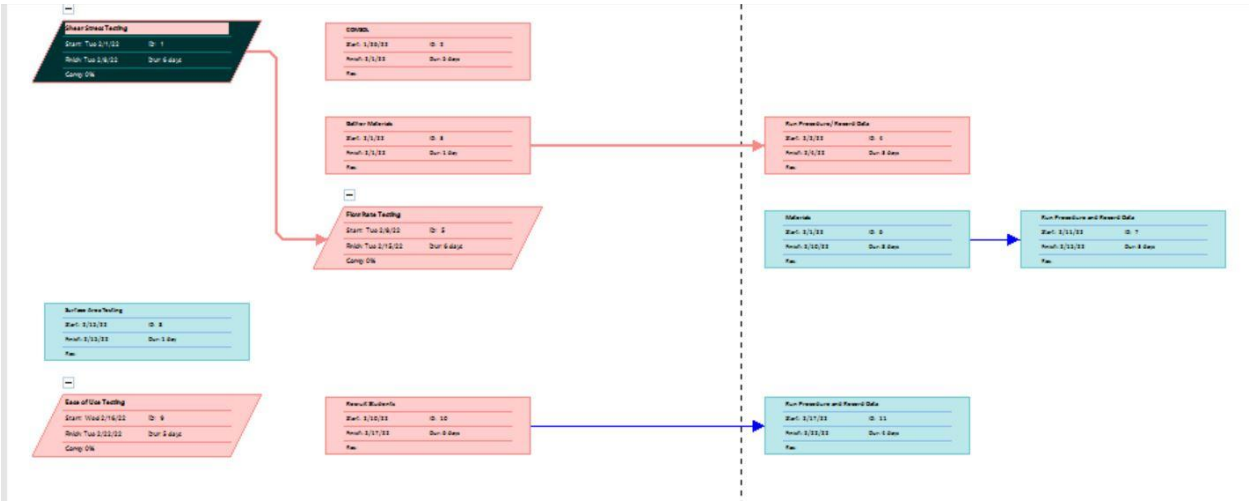


*Figure 32: Test Plans Network Diagram.*

| Task # | Task | Resources | Time (days) | Start Date | End Date |
|---|---|---|---|---|---|
| 1 | **Shear Stress Testing** | | 7 | 2/1 | 2/8 |
| 2 | Perform COMSOL Simulation | Computer access with COMSOL for long duration, 2 team members | 7 | 1/26 | 2/1 |
| 3 | Gather Materials | MPS Lab students assistance | 1 | 2/1 | 2/1 |
| 4 | Run Procedure & Record Data | Microbeads, camera with known framerate, working prototype | 4 | 2/4 | 2/8 |
| 5 | **Flow Rate Testing** | | 7 | 2/8 | 2/15 |
| 6 | Buy and Acquire Materials | Computer access, access to BMED office, MPS Lab students assistance | 10 | 2/1 | 2/10 |
| 7 | Run Procedure & Record Data | Miligram scale, beaker, water, 3 team members present | 5 | 2/10 | 2/14 |
| 8 | **Surface Area Testing** | Measuring tape (cm), access to Room 329 | 1 | 2/15 | 2/15 |
| 9 | **Ease of Use Testing** | Finished protoype, Operations Manual | 6 | 2/22 | 2/22 |

*Figure 33: Table organizing our tests, their subtasks, planned dates of execution, and resources needed for each step in our test plan.*

Shear Stress Testing

Because both the COMSOL aspect and physical aspect of the shear stress testing is time and resource demanding, we want to limit this testing to 2 trials. Planning on using a two-sample t-test, we ran a power analysis with a difference of 0.05, and a predicted standard deviation of the two trials 0.01. The resulting power for this test is 72%. Below are our null and alternative hypotheses for this testing.

H0: There is no significant difference between the shear stress found in our Comsol model and our physical testing of shear stress.
H1: There is a significant difference between the shear stress found in our Comsol model and our physical testing of shear stress.

*Protocol:*

1. Load SolidWorks model of microfluidic device into Comsol.
2. Run a parametric sweep of flow rates through the device at the point close to the wall at the input of the device to get resulting velocity of the medium through the microfluidic device. Use a temperature of 37 C.
3. Gather velocity data.
4. Place microbeads in a solution of water.
5. Remove chassis lid and remove the syringe from adapters.
6. Prime the device with 96% ethanol to get rid of bubbles.

7. Using the LabSmith Micro Particle Image Velocimetry microscope, find the wall of the microfluidic device by the inlet and focus in on the bottom surface as accurately as possible.
8. Fill the syringe with the water mixed with the microbeads.
9. Place the filled syringe back into the adapters of the pump.
10. Set the parameters to a flow rate of 75.8 uL/hr. Repeat these steps with the parametric sweep of flow rates inputted into COMSOL.
11. Record velocities for approximately 10 seconds.
12. Using the Excel data that the microscope produces, find the velocity of the microbeads at each of the flow rates by taking an average of the velocity magnitude over the time it was recorded.
13. Perform a two-sample t-test to compare the average velocity found in the Comsol trials to the average velocity found in the physical testing.

We decided to run a parametric sweep of flow rates, starting with a flow rate of 75.8 and decreasing by a factor of 10 to the flow rate that the pump will be typically working under, 0.76 uL/hr.

*Sample Size:*
Our target sample size is simply 2 due to the resources available to us.

*Expected Results:*
We expect that the physical testing of our shear stress will produce the same shear stress found in Comsol at a flow rate of 0.76 uL/hr. Because velocity is directly proportional to shear stress, we will be comparing velocity of the analytical and physical models. The expected velocity at a 0.76 uL/hr flow rate is approximately 3.60 um/s.

*Pass/ Fail Criteria:*
To pass the test, there must be no significant difference between the physical and analytical velocities, using an alpha of 0.05.

*Contingency if Testing Does Not Meet Criteria:*
If the pump does not deliver the shear stress criteria, we will re-examine each interface of the device from the motor to the contact of the pusher of the syringe and redesign them so that the precision of the speed is not lost in the interfaces.

*Materials Needed:*
To run this test, we will need the working prototype, a lab space for an extended amount of time, microbeads, LabSmith Micro Particle Image Velocimetry microscope, the COMSOL application on a computer, and at least 2 personnel present at every point.

Flow Rate Testing

Planning on using a one-sample t-test, we ran a power analysis with a target power of 80%, a difference of 0.03, and a predicted standard deviation of 0.025. Through that power analysis we concluded that we need 3 trials for our sample size for flow rate testing.

H0: There is no significant difference between our target flow rates and the average of the flow rates found in the 3 trials.
H1: There is a significant difference between our target flow rates and the average of the flow rates found in the 3 trials.

*Protocol:*

1. Place a plastic beaker on the Scientech digital milligram scale and zero out the weight of the beaker.
2. Open the chassis and remove the syringe from the adapters.
3. Place 1 mL of water into the syringe.
4. Secure syringe back into the adapters and close chassis lid.
5. Set flow rate to 28,800 uL/hr
6. Set time to 50 seconds.
7. Record weight of water accumulated in the beaker in those 50 seconds.
8. Calculate flow rate with knowledge of density of water and record data points.
9. Run the test 3 times at each flow rate, getting progressively smaller until the measurement becomes significantly inaccurate.
10. Use a sensitive gauge to test displacement for the lower slow rates to ensure correct linear speed being delivered to the syringe to get desired flow rate.
11. Create a graph that relates the physical displacement to the flow rate found via testing on the scale.
12. Interpolate the smaller flow rate using this relation.
13. Perform a one-sample t-test to compare the mean of the 3 interpolated data points to the target flow rate of 0.76 uL/hour.

We decided to run a parametric sweep of flow rates, starting with a group of fairly high ones in order to get a solid trend of how our pump acts at these higher speeds in order to interpolate to data at the lower flow rates. Thus, we chose to start with an input delay of 1 ms, correlating to a flow rate of 28,800 uL/hr, and decrease slowly initially to 9600 uL/hr to 1920 uL/hr, the to 758 uL/hr where we then decreased the flow rates by a magnitude of 10 until we hit the main flow rate that it will be operating under, 0.76 uL/hr.

Using the flow rate we inputted into our device, we calculated the expected mass of water we should get on the scale after a set amount of time using the following equation:

$$Flow\ Rate\ (ml/hr)\ *\ time(hr)\ =\ Volume\ (ml)\ *\ 1g/1ml\ =\ mass\ (g) \quad \textbf{(2)}$$

In order to use the mass we actually got to calculate the flow rate that was actually being delivered, we used the following equation:

$$(Mass\ (g)/time(hr))\ *\ 1000\ =\ Flow\ Rate\ (uL/hr) \quad \textbf{(3)}$$

*Sample Size:*
Perform this same test 3 times to track both accuracy and precision.

*Expected Results:*
We expect that after calculating the flow rate delivered based on the mass measured by the scale over a set amount of time, the % error between the flow rate inputted and flow rate delivered will be 15% or less.

*Pass/ Fail Criteria:*
To pass the test, the average of the 3 trials must be within a 15% error. Anything outside of this range we will consider failure.

*Contingency if Testing Does Not Meet Criteria:*
If the pump fails the flow rate criteria, we will re-examine each interface of the device from the motor to the contact of the pusher of the syringe and redesign them so that the precision of the speed is not lost in the interfaces.

*Materials Needed:*
The materials needed include a plastic beaker, Botany digital milligram scale, a 1 ml syringe, tubing, and water. All members must be present.

<u>Surface Area Testing</u>
H0: There is no significant difference between our measured surface area and our maximum possible surface area.
H1: There is a significant difference between our measured surface area and our maximum possible surface area.

*Protocol:*
1. Measure surface area of the bottom of the pump, including the electrical components, with a ruler in cm.
2. Place the device in allotted space.

3. Turn on running water and see if it is in reach of the electrical components of our pump.

*Sample Size:*
We will only need 1 data point.

*Expected Results:*
With our design we expect the surface area to be 425 cm^2.

*Pass/ Fail Criteria*:
It passes if it fits in the allotted space that is 695 cm^2 big and electrical components are not in reach of the water.

*Contingency if Testing Does Not Meet Criteria*:
If the syringe pump does not fit on the counter, we have to consider redesigning the chassis so that it is small enough to fit in the desired location. We can also look at the placement of the electrical components and rearrange them if they are taking up too much surface area.

*Materials Needed*:
For this test we only need a measuring utensil, preferably a measuring tape, and we need one member present when conducting

<u>Ease of Use Testing</u>
H0: It will take five or less steps to run the device from power-on to running.
H1: it will take more than five steps to run the device from power-on to running.

*Protocol:*
1. Number the steps of the Operation's Manual, only including the actual directions and not the safety notes.
2. Count how many steps result.
3. Verify if it is 5 or less.

*Sample Size:*
We only need 1 sample of data to confirm that the number of steps is 5 or less.

*Expected Results:*
We are expecting that our pump can be operated in approximately 4 steps.

*Pass/ Fail Criteria:*

We will consider over 6 steps a failure. Anything less than 6 passes.

*Contingency if Testing Does Not Meet Criteria:*
We will look into different keypads as well as different types of screens. For this test the only materials we will need are the instructions printed on paper and our computers to take notes during each test run. For each test we will have two of our three group members present to make sure that all steps are counted and any feedback from participants is noted.

*Materials Needed:*
One member of our team is needed for this test, otherwise no other materials are needed.

*Note: We updated the protocol for both the shear stress and flow rate testing as we began testing and started discovering some extra steps that needed to be implemented and extra materials needed.*

**Testing Data and Analyses**

Shear Stress Testing

*Results*

The following table displays the velocities acquired from our physical testing via the LabSmith Micro Particle Image Velocimetry microscope as well as the velocities found in COMSOL via the analytical model of our microfluidic device. Because shear stress is directly proportional to velocity, we decided to directly compare velocities in our physical and analytical models to determine the accuracy of shear stress delivered by our device. We made the assumption that our COMSOL analytical model is accurate, so we then implemented physical testing to compare the velocities (and thus shear stresses) that we obtain to these values, and the table below were the results.

**Table 12: Shear Stress Testing Data**

| Flow Rate (uL/hr) | Velocity (um/s) | COMSOL Velocity (um/s) | %Error |
|---|---|---|---|
| 75.8 | 295.6 | 358.91 | 17.6 |
| 7.6 | 102.11 | 35.99 | 183.7 |
| 0.76 | 69.5 | 3.60 | 1830.5 |

We ran a two-sample t-test comparing the velocities and thus shear stresses of the analytical and physical model, and unfortunately the difference was significant using an alpha of 0.05.

*Analysis*

There are many explanations for the significant difference between the velocities in our physical and COMSOL models. First, the microfluidic devices that we used for the physical testing were slightly different geometries than the model that we tested in COMSOL, affecting the speed the fluid travels through the specific point we were examining and thus affecting the shear stress felt by the cells in those areas. This was simply due to the resources that we had at hand. Some of the variation in shear stresses between our physical and analytical model can also be attributed to interface compliance, as our pusher block was not securely fastened when we performed the physical testing. We have since fastened the block securely via screws, which we can assume will only improve the accuracy of the shear stress our pump delivers. Due to lack of time and resources, we were unable to re-run the testing after we had fixed this compliance.

Additionally, the testing via the LabSmith Micro Particle Image Velocimetry microscope introduced some inaccuracies, as the measurements were tainted any time a bubble was introduced into the microfluidic device. Despite best efforts to remove such bubbles, they still infiltrated the device during our testing. Additionally, the way that our pump delivered the fluid via small pulses at the lower speeds caused a nonsteady flow through the microscope. This caused significantly inconsistent velocities over the time period we recorded, and we averaged those values to find the velocity at that inputted shear stress. Ultimately, we can appreciate that with the decrease in flow rate input, the velocity in the device and thus shear stress felt by the cells decreased as well, confirming our pump delivers shear stress values based on the flow rate inputted by the user. In the future, we would like to run more trials, and be able to run each flow rate for longer at a time before recording so that the speed through the microfluidic device really stabilizes.

<u>Flow Rate Testing</u>

*Results*

We started by testing the displacement of the pusher block over time to ensure that the correct speed was being delivered based on the input of the user. We decided to start with this so that any further error in our flow rate testing could be attributed to variation in our testing procedure as well as the interfaces of the device, specifically the pusher block and syringe base holder adapters. The following were the results of this physical gauge testing when running each flow rate for 5 minutes.

**Table 13: Physical Displacement Testing Data**

| Flow Rate (uL/hr) | Predicted Displacement (mm) | Displacement (mm) | Percent Error |
|---|---|---|---|
| 28,800 | 116.7 | 114.3 | 2.06 |
| 9600 | 39.8 | 41 | 3.02 |
| 1920 | 7.98 | 8.12 | 1.72 |
| 758 | 3.16 | 3.19 | 0.95 |
| 75.8 | 0.316 | 0.32 | 1.27 |
| 7.6 | 0.0317 | 0.0315 | 0.63 |
| 0.76 | 0.003175 | 0.003075 | 3.15 |

After ensuring that our motor delivers accurate displacements, we proceeded to test the flow rates using the scale method as described in our test plans. Equations 2 and 3 found in the test plan section were used to calculate flow rate. The following are our results from that testing.

**Table 14:Flow Rate Testing Data for First Trial**

| Flow Rate Inputted (uL/hr) | Average Tested Flow Rate Based on Mass of Water Acquired (uL/hr) | Displacement of Block at Flow Rate Inputted(mm) |
|---|---|---|
| 28,800 | 25,835 | 114.3 |
| 9600 | 8201 | 41 |
| 1920 | 1513 | 8.12 |
| 758 | 489 | 3.19 |
| 75.8 | 40.843 | 0.32 |
| 7.6 | 3.209 | 0.0315 |
| 0.76 | 0.250 | 0.003075 |

We graphed flow rate versus displacement at that flow rate in order to interpolate the smaller flow rates using displacement after 5 minutes as our x-axis, as these smaller flow rates were near impossible to test accurately because of the sensitivity and

variation of the scale. A power curve matched our data the best, so we implemented this formula seen on the graph below to interpolate.

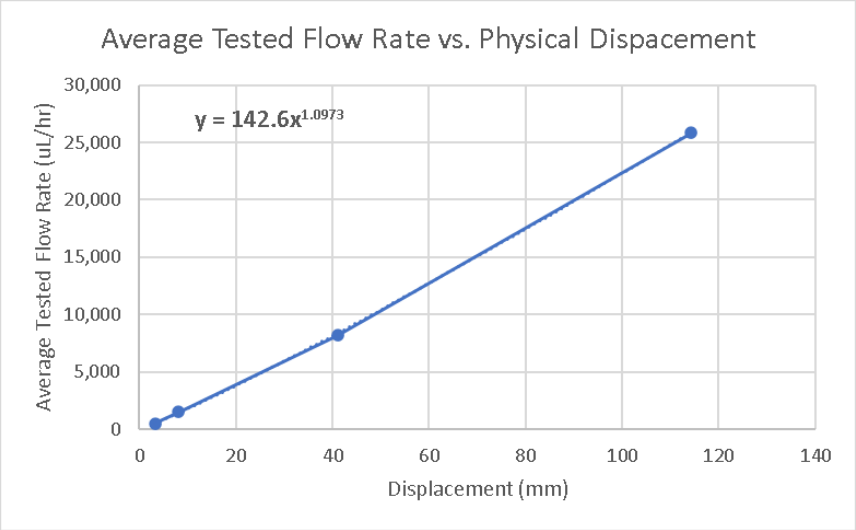



*Figure 34: Power fit to our average flow rate versus physical displacement data used for interpolation of smaller flow rates.*

We had assumed a lot of our error to be due to both the variation due to the testing procedure as well as due to our interfaces not being securely fastened. Once we updated the way that our adapters were fastened to the motor, we wanted to perform testing once more to show the improvement made. These are the results from this second round of testing.

**Table 15: Flow Rate Testing Data for Updated Design**

| Correlated Flow Rate (uL/hr) | Updated Tested Flow Rate Based on Mass of Water Acquired (uL/hr) | %Error | Displacement (mm) |
|---|---|---|---|
| 28,800 | 25471.0 | 11.6 | 114.3 |
| 9600 | 8300.0 | 13.5 | 41 |
| 1920 | 1392.7 | 27.5 | 8.12 |
| 758 | 512.0 | 32.5 | 3.19 |
| 75.8 | 41.047 | 45.9 | 0.32 |
| 7.6 | 3.252 | 57.2 | 0.0315 |
| 0.76 | 0.255 | 76 | 0.003075 |

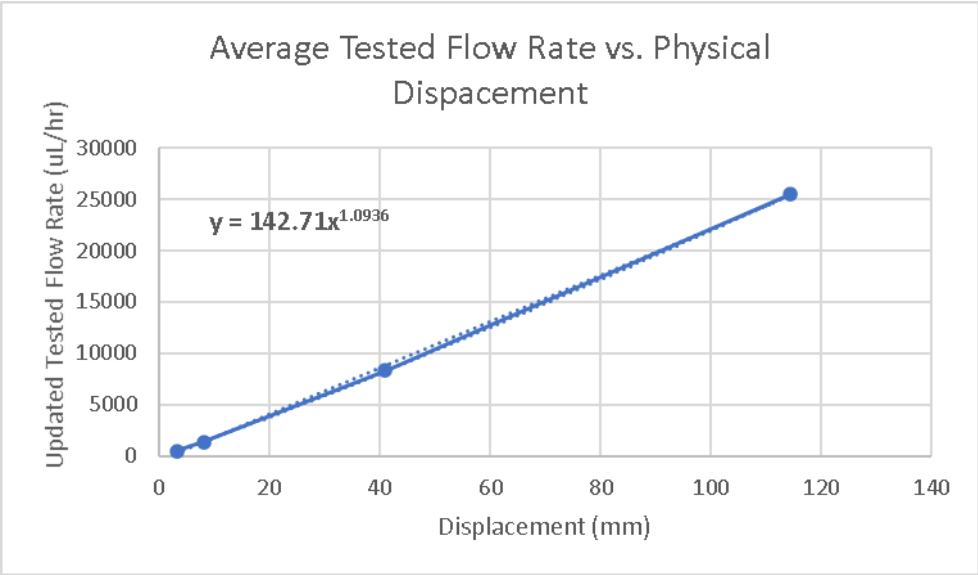



*Figure 35: Power fit to the flow rate found in our second round of testing versus physical displacement data used for interpolation of smaller flow rates.*

Using the flow rate data above, we ran a 1-sample t-test comparing the inputted flow rate of 0.76 uL/hr to the interpolated value of what our pump performs, and there was a significant difference in these values when using an alpha of 0.05.

*Analysis*

Overall, our tests resulted in values that were precise, hitting the same output values each time, but not accurate to the flow rate we expected it to be. The first time running the trials, we attributed this potential variation to the poor adhesive between the pusher block and its adaptor. In order to correct this, we used hardware to screw the adapter to the top of the pusher block, and used epoxy to adhere the other adapter to the end of the device. After the improvement for the attachments, we ran the tests over and ended with very similar results. This lack of change in results after the hardware attachments suggests that the variation is not in our device. The pump is delivering the same outputs each time, which is what we were looking to test. However, these outputs do not closely match what they are programmed to be, especially at the lower flow rates. We can conclude that the variability arises in the system somewhere between the syringe needle, the tubing, the plastic beaker, and the scale. Thus, our pump might not necessarily be the culprit of this variation. A way to figure this out in the future if given more time would be to measure the flow rate right as it is leaving the tube via an alternative method.

<u>Surface Area Testing</u>

*Results*

Below is a table summarizing our specification for surface area, what we originally expected our surface area to be when writing our test plans, and our final measured surface area of the device.

**Table 16: Surface Area Testing Data**

| Surface Area Specification | Expected Surface Area | Measured Surface Area |
|---|---|---|
| <695 ㎠ | 425 ㎠ | 431.85 ㎠ |

*Analysis*

We were able to meet the specification on surface area so that our pump would be able to fit on the desired location above the microfluidic environment. We simply measured the bottom of our chassis, which is what the elements of our pump all reside in.

Ease of Use Testing

*Results*

Following our testing procedure, we numbered the steps in our operations manual, not including the safety warnings, as follows.

1. When the pump is not operating, open the lid of the chassis to remove the syringe from the adapters and fill with desired fluid.
2. Place the loaded syringe back into the holes of the adapters and close the lid of the chassis.
3. Insert desired flow rate via the buttons.
4. Turn on the power supply and briefly watch the movement of the pump and the syringe to ensure parts are securely fastened and working properly.
5. Turn off power to the syringe once done using.

Below is a table summarizing our specification for ease of use, what we originally expected our number of steps to be when writing our test plans, and our final measured number of steps of our final prototype.

**Table 17: Ease of Use Testing Data**

| Ease of Use Specification (Max # of Steps) | Expected Amount of Steps | Measured Amount of Steps |
|---|---|---|
| 5 ±1 | 4 | 5 |

*Analysis*

We can conclude that our syringe has met our specification for ease of use. In general, the directions are fairly straight forward and should be able to be performed by both the users in the MPS lab as well as any student on campus.

**Conclusions**

After analyzing the data from our volumetric flow rate testing, we found that the flow rate our device was delivering was inconsistent with the flow rate that we expected it to create. At the higher flow rates that we tested, the amount of liquid dispensed was routinely and consistently less than the amount of liquid that we had calculated to be dispensed. Additionally, as our flow rate became smaller and smaller, the testing procedures we had developed proved to be insufficient for collecting accurate results about microfluidic flow rates. Plotting displacement versus flow rate and interpolating for smaller flow rates helped us create a profile of flow rate. Ultimately, our syringe pump is highly precise, able to replicate flow rates that are close to our desired flow rate. However, an interfacing issue between our syringe pump and the testing equipment limits our ability to accurately test our device. Thus, we can conclude that we were unable to hit the specification for our flow rate.

Our shear stress testing was not able to provide our group with much useful information. Only one of our values for shear stress came within 15% of our expected shear stress value. For our shear stress testing, we ran into a plethora of issues. There were several instances in which we tried to conduct our shear stress testing, and events outside of our control prevented us from completing our testing (broken microfluidic device, lack of time, air bubbles appearing in the microfluidic device, etc.). Additionally, we calculated and compared shear stress values for our test and our comsol value assuming fluid flow between parallel plates taken at the midpoint between the two points. In reality, it was impossible for us to tell where our microscope objective was focusing. We tried to get the objective focused as closely to the midpoint as possible, but there was no way for us to know for sure. Air bubbles consistently appeared in our microfluidic device, feeling like they appeared out of nowhere. This limited fluid flow and in some cases accelerated localized fluid flow as the cross sectional area of the device where fluid could flow was decreased. Ultimately, due to all the issues we ran into, we were unable to validate our specifications for shear stress.

Our human interfacing and manufacturing testing proved to be more successful. Our surface area specification of 695㎠ was validated, as the final surface area for the device was measured out to be 431.85㎠. For our ease of use testing, we developed an initial specification of 5 +/- 1 step for initial use. We also validated this specification, achieving 5 steps necessary for initial use.

**Discussion**

After completing our tests, we concluded that we met two of the four specs. The flow rate and shear stress testing did not provide the results we were looking for. One potential solution was the attachment method of the adaptors. We had both of the adaptors attached with duct tape, which gave the possibility for give when the pusher block moves. In order to attempt to solve our problem, we used hardware to screw the top adapter onto the pusher block, and we used epoxy to attach the other adapter to the end of the device. After re-running the flow rate testing with our new iteration, we found that the results were not significantly different from the test results from the duct tape adaptors. We concluded that the adhesive methods were not the problem, and there must have been variance between the syringe, the tubing, the beaker, and the scale.

Unfortunately, our time is over and we have done all that we can on our syringe pump. However, there is so much more we would like to change if we had more time and resources. First of all, we would like to re-run our flow rate and shear stress testing. We found that the data was precise but not accurate; we concluded that there was variance between the syringe, the tubing, the beaker, and the scale. For the new flow rate test, we would need higher quality measuring utensils that provided less variance. Our scale was difficult to zero and would often waver even after there was no water falling onto it. If we could adjust the flow rate testing and solve the issue in the interface between parts, we would then be able to re-run the shear stress testing and would potentially get better results. In the re-run of the shear stress testing, we would need better training on how to use the LabSmith Micro Particle Image Velocimetry microscope so that we were not reliant on Dr. Hawkins to help us every time there was a bubble in the microfluidic device. The bubbles were a common problem that may have caused variance, and it was something that our team did not have the ability to prevent. Overall, our next steps would be pouring more time and resources into fixing the inaccuracy of the flow rate and shear stress delivered by our device before delivering it to the lab.

Another aspect of the syringe pump that we completed but would improve if given the time and the funds is the user interface. We were able to program buttons and a screen that are very simple. With more time and funds, we would be able to program a more sophisticated keypad and screen. This would allow for the user to program the pump to run for a duration rather than forcing the user to turn the device off when it is done running. Another button we would like to add is a power button; this is because the pump is currently turned on and off with the power switch on the power strip it is plugged into.

Of our four specifications, we are content with the results from two. We met the criteria for the surface area testing, making the syringe pump more compact and easy to move than the existing pump in the microfluidics lab. We are also happy with our ease of programming specification; we were able to make the pump easy to use, being able

to program from start to finish in five steps .There is nothing that we would change in order to improve these two specs.

**Acknowledgements**

We would like to thank Dr. Heylman as our sponsor for entrusting us with this project. Each of us members have grown significantly in our knowledge of the design process and in our skills as engineers through developing our syringe pump. Additionally, we would like to thank Dr. Hawkins for providing help and resources for testing of our shear stress.

**Appendix A**
Arduino Sketch Code

```
#include <LiquidCrystal.h>
const int rs = 7, en = 6, d4 = 2, d5 = 3, d6 = 4, d7 = 5;
LiquidCrystal lcd(rs, en, d4, d5, d6, d7);
//Input & Button Logic
const int numOfInputs = 4;
const int inputPins[numOfInputs] = {10,11,12,13};
int inputState[numOfInputs];
int lastInputState[numOfInputs] = {LOW,LOW,LOW,LOW};
bool inputFlags[numOfInputs] = {LOW,LOW,LOW,LOW};
long lastDebounceTime[numOfInputs] = {0,0,0,0};
long debounceDelay = 5;


//LCD Menu Logic
const int numOfScreens = 1;
int currentScreen = 0;
String screens[numOfScreens][2] = {{"Flow Rate","uL/hr"}};
int parameters[numOfScreens];
String parametersUnits[numOfScreens] = {"uL/hr"};


void setup() {
  for(int i = 0; i < numOfInputs; i++) {
    pinMode(inputPins[i], INPUT);
    digitalWrite(inputPins[i], HIGH); // pull-up 20k
  }
  //Serial.begin(9600);
  lcd.begin(16, 2);

pinMode(8,OUTPUT);
pinMode(9,OUTPUT);
digitalWrite(9,LOW); // LOW to bring the slide back and HIGH to push the slide forward
```

```
}



void loop() {
//   put your main code here, to run repeatedly:

  long pulsedelay;
  pulsedelay = 2880000000 / parameters[currentScreen];
  digitalWrite(8, LOW);
  digitalWrite(8,HIGH);
  delayMicroseconds(pulsedelay);
// edit the above delay to modify the output flow rate of the device,

  setInputFlags();
  resolveInputFlags();
}

void setInputFlags() {
  for(int i = 0; i < numOfInputs; i++) {
    int reading = digitalRead(inputPins[i]);
    if (reading != lastInputState[i]) {
      lastDebounceTime[i] = millis();
    }
    if ((millis() - lastDebounceTime[i]) > debounceDelay) {
      if (reading != inputState[i]) {
        inputState[i] = reading;
        if (inputState[i] == HIGH) {
          inputFlags[i] = HIGH;
        }
      }
    }
    lastInputState[i] = reading;
  }
}

void resolveInputFlags() {
  for(int i = 0; i < numOfInputs; i++) {
    if(inputFlags[i] == HIGH) {
      inputAction(i);
```

```
      inputFlags[i] = LOW;
      printScreen();
    }
  }
}

void inputAction(int input) {
  if(input == 0) {
    if (currentScreen == 0) {
      currentScreen = numOfScreens-1;
    }else{
      currentScreen--;
    }
  }else if(input == 1) {
    if (currentScreen == numOfScreens-1) {
      currentScreen = 0;
    }else{
      currentScreen++;
    }
  }else if(input == 2) {
    parameterChange(0);
  }else if(input == 3) {
    parameterChange(1);
  }
}

void parameterChange(int key) {
  if(key == 0) {
    parameters[currentScreen]++;
  }else if(key == 1) {
    parameters[currentScreen]--;
  }
}




void printScreen() {
  lcd.clear();
  lcd.print(screens[currentScreen][0]);
```

```
  lcd.setCursor(0,1);
  lcd.print(parameters[currentScreen]);
  lcd.print(" ");
  lcd.print(screens[currentScreen][1]);
}
```

**Appendix B**

Rudimentary Arduino Sketch

```
void setup() {
  // put your setup code here, to run once:
pinMode(8,OUTPUT);
pinMode(9,OUTPUT);
digitalWrite(9,HIGH); // LOW to bring the slide back and HIGH to push the slide forward
}

void loop() {
  // put your main code here, to run repeatedly:
digitalWrite(8, LOW);
digitalWrite(8,HIGH);
delay(3);
// edit the above delay to modify the output flow rate of the device,
}
```