An Open-Source Syringe Pump for Anesthesia Delivery in Developing Countries

a report done for

UNIVERSITY OF NEBRASKA MEDICAL SCHOOL
&
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by

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Executive Summary:

The state of medical technology in developing countries is dire. There is a scarcity of devices that have the potential to save lives, and of the available devices, they are often not utilized because they are broken, or the hospitals lack the expertise to reliably use them.
Working with both the Utah School of Medicine and the University of Nebraska Medical School, we hope to minorly alleviate this issue. The goal of this project is to design a 3d printable syringe pump for application specifically in developing countries.

This quarter we have made major progress towards this goal. Primarily, we have focused on refining the design we selected last quarter and running experiments to quantitatively measure the effectiveness of this device against the technical specifications selected last quarter.

The selected design's actuation is designed around a lead-screw system that pushes the syringe forward proportional to how fast a stepper motor rotates. This decision was based on the reliability and affordability of this method of actuation. The actuation method is by far the most complex component of this device, so the rest of the device has been designed around the actuator. A CAD model and current prototype of this device are shown in the figure below.

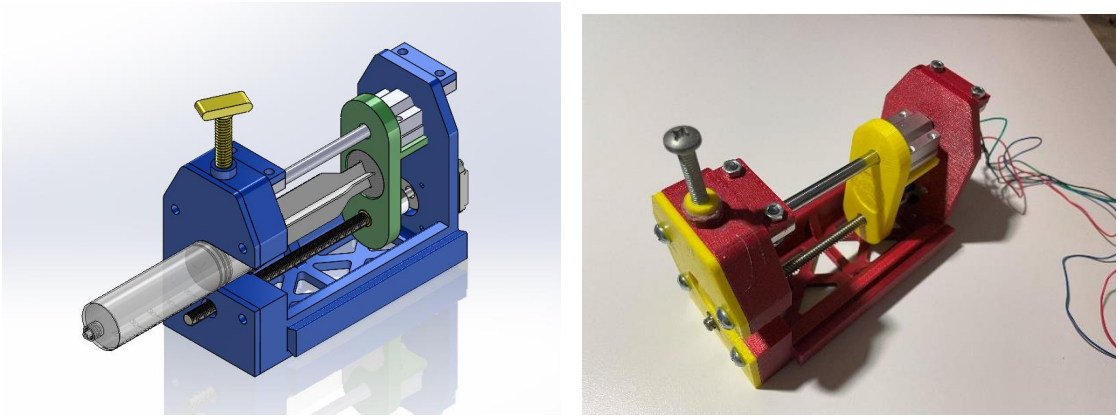


Figure 1: CAD model(left) and physical model (right) of current prototype.

At the current state, the team has verified that the flowrate technical specifications should be met based on the experiments that will be outlined later in this report. In addition, the design is greatly reduced in bulk compared to the design presented in the Quarter 1 report.

Moving forward, the team will be continuing to focus on weight and size reduction by running more FEA simulations to determine areas when less support is necessary, as well as running experiments to verify the devices success for the currently untested technical specifications.

Disclaimer:

The contents of this report were prepared by senior mechanical engineering students at Rose-Hulman Institute of Technology. We feel confident in our work as students. However, all material should be reviewed by an appropriate professional before implementation.

Table of Contents:

Problem Definition:

Problem Statement:
Design an open source, syringe pump to be additively manufactured, repaired, and operated in developing countries.

Technical Challenges:
There are many needless medical mortalities in developing countries. These mortalities come from a variety of ailments, and because of this, no one device can act as a panacea. These preventable mortalities largely arise from a lack of access to medical technology and trained professionals.

In cooperation with the Utah School of Medicine, the University of Nebraska Medical School hopes to improve conditions by designing medical technology specifically for application in developing countries. Particularly, they are interested in designing a syringe pump because of its versatility.

Syringe pumps can deliver or pull continuous dosages of medications for a wide range of applications. For instance, they can be used for IV pulls, epidurals, assorted catheters, and insulin pumps. These applications vary in importance, but for some instances (pre-eclampsia, post-partum hemorrhage, etc.), the ease of access to syringe pumps is a direct determinant of patient outcomes.

The syringe pump market in the United States is dominated by the BD Alaris Syringe Pump, shown in 3, this pump is designed for operation in a modern Operating Room (OR). By extent, this means that the syringe pump market in developing countries is also dependent on this pump, approximately 95% of medical equipment used in developing countries is imported from developed countries [1]. The Alaris Pump is not an optimal pump for developing countries, this leaves a niche available in the syringe pump market.

Figure 1: BD Alaris pump, the industry standard syringe pump in developed countries.

However, designing medical technology for use in a developing country presents a handful of difficult technical challenges. Firstly, to reduce lead times and shipping costs, all parts need to be manufactured locally or purchasable from 'off-the-shelf' dealers. To ensure that all hospitals can easily manufacture these parts, we are limiting our manufacturing methods to a 3D printer. Secondly, the device will likely not always be operated by someone who is well trained with the technology. This means that the device needs to be intuitive and simplistic. Thirdly, the device needs to work accurately in non-ideal operating conditions, this will come in the form of backup power supplies and a rugged design.

These challenges are combined with all the difficulties associated with any medical device. The device needs to be very accurate to be useful. Also, it must alert the operator whenever something unexpected has happened. With so many challenges in mind, we are collaborating with a sibling team, the Electric and Computer Engineering (ECE) team, where they will focus on the electronic power supplies and UI aspects on the design. The role of our team would be to design this device, process its outputs and determine if anything abnormal is occurring while the device is in use.

It is important to note that the pump we are designing is not ever going to be used for medical use. For this reason, getting this pump approved by the FDA is outside of the scope of the project. Instead, we are designing the pump exclusively for lab use, however we are still aiming to create a product that will not require a total redesign in order to get FDA approved.

Project Needs:

The following list breaks down the project needs that we have identified. They are generally ranked from most important to least important, though each of these is considered a necessity for this device to be considered successful.

1. The device should output an accurate flow rate.
    a. Flow rate should be adjustable.
    b. Flow rate should be consistent through a wide range of values.
2. The device must be safe.
    a. High accuracy for the volumetric flow rate.
    b. Flow rate should not jam.
    c. Should alert users when unexpected events occur.
    d. Can be sanitized easily.
3. The device should be simple to build.
    a. Components of the device should be easily replaceable.
    b. Components should be able to be assembled and maintained with simple tools.
4. The device should be modular.
    a. Ability to add additional syringe pumps.
    b. Information should be able to be shared between modules.
5. The system must be cost-effective.
6. The device must be simple to operate.
7. The device should be primarily manufactured by 3D printing and 'off the shelf' components.
8. The system can have one translational degree of freedom.
9. The device should mount to an IV pole.
10. The device output should remain constant with fluctuations in the environment.
    a. External temperature should not affect flow rate.
    b. Electronics should be shielded from external elements.
11. The device should not be cumbersome.
    a. The device should be small and lightweight.

These project needs will be used to guide our design and decision-making processes as we continue with this project. These will appear primarily in the technical specifications portion of the document but were also used during the concept selection process.

As we are collaborating with another Senior Design Team, we developed a functional model to better define the device functions that will be covered by each team. The functional model below captures the project functions we have identified in the section above.

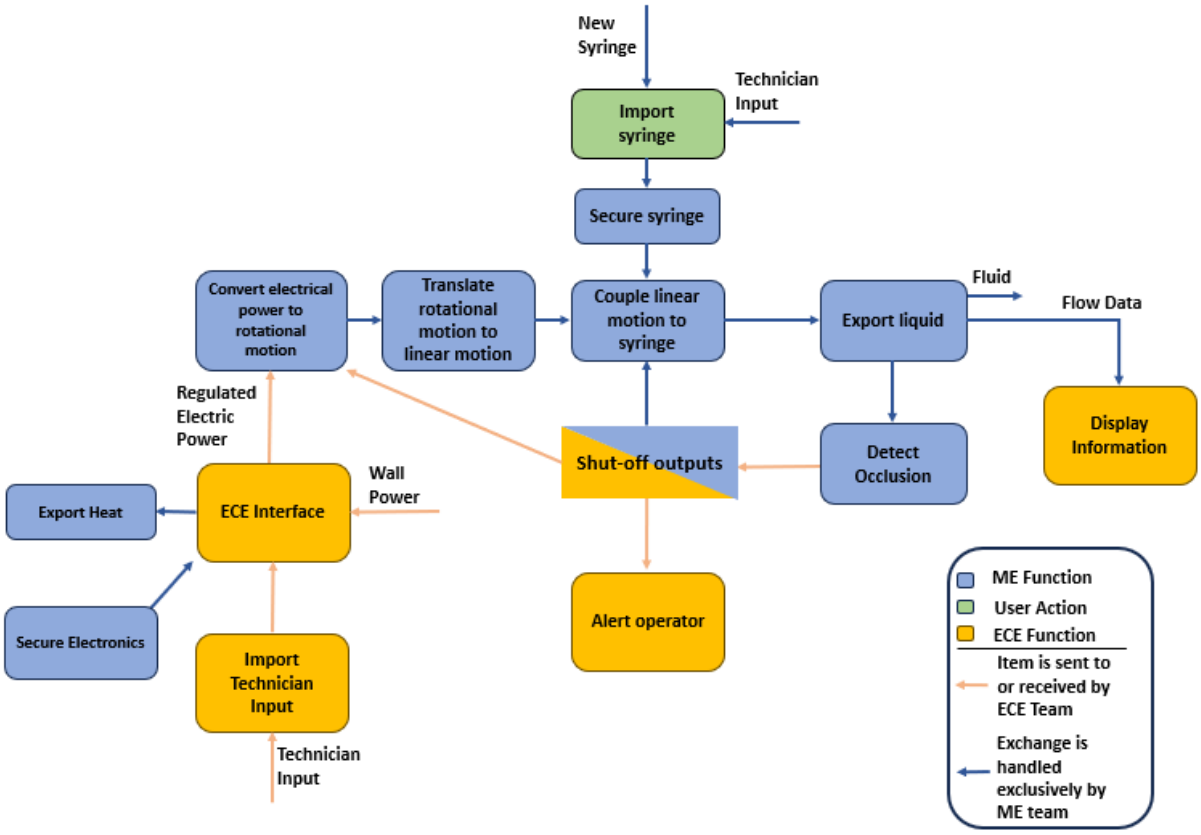


Figure 2: Functional model of syringe pump.

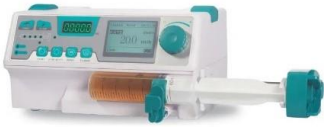
This project interfaces closely with an ECE sibling team. For this reason, we have indicated design functions that will be handled by them with a different color. However, a lot of these functions will be reliant on our outputs and vice versa. To account for this, we have each selected one team member to regularly communicate with another to allow for rapid communication.

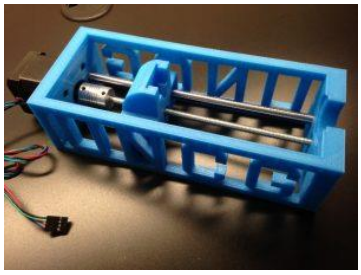
Benchmarking

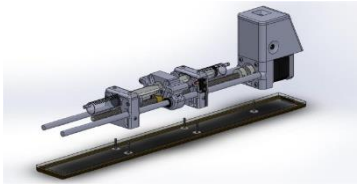
Benchmarking was a critical part in identifying the products that are currently being used in the professional world. The syringe pumps used in the professional world fall into two categories: medical use and research use. The pumps used in the medical field are typically far more expensive than research use pumps. Since we are designing a pump that is currently utilized for research, but that will ultimately be used as a medical device, we thought it was worthwhile to examine both fields.

Table 1: Benchmarked Products

Product Name (all names linked)	Description	
 Perfusor Space	**Quantitative Analysis**	Accuracy: ±2% Weight (lb): 3.1 Dimensions (in): 9.8×2.6×5.9 Cost ($): 1,650.00 Medical grade (y/n): yes
	Verbal Analysis	- Possesses a multitude of safety systems: automatic syringe drive arm and piston brake to prevent inadvertent bolus during syringe change. - Allows for both weight-based dosing limits and the creation of patient profiles. - Utilizes non-numeric keypad to reduce manual programming errors, intentionally simple interface to reduce training time. - Contains pre-programmed 'drug library'. - Built in alarm is programmable to control volume levels.
 Medfusion 3500 Syringe Pump	**Quantitative Analysis**	Accuracy: ±2% Weight (lb): 3.8 Dimensions (in): 5.1×10.5×5.5 Cost ($): 1490.00 Medical grade (y/n): yes
	Verbal Analysis	- Allows for programmable dose limits. Dose can be delivered through assorted methods, including: $\frac{mL}{hr}$ body weight/mass, volume over time, custom dilution, and intermittent. - Safety features include occlusion detection, post-occlusion bolus reduction, 'smart' safety limits. If any of these are tripped, they alert the physician. - Capable of up to 10 hours of battery capacity. - MRI compatible. Capable of being used in an MRI room with a magnetic field of up to 150 Gauss. - Accepts all syringe sizes cc-60cc. - Designed to mount to IV pole.

Product Name (all names linked)	Description	
 InfusionONE Single Channel Syringe Pump	**Quantitative Analysis**	Accuracy: ±1% Weight (lb): 3.8 Dimensions (in): 5.75×8.75 Cost ($): 805.00 Medical grade (y/n): no
	Verbal Analysis	- This syringe pump hold utilizes a single syringe. It can hold up to 60 mL of fluid. It features an easy-to-use keypad that can be used to autonomously control the syringe pump system. - It can be controlled independently or can be hooked up to a computer. - It is capable of both infusion and withdrawal. - It has multiple safety features, such as stall detection. - The syringe is pushed with a force of about 15 lbs at its top speed and 35 lbs at its lowest speed.
 MD910 Medical Syringe Pump	**Quantitative Analysis**	Accuracy: ±3% Weight (lb): 4.85 Dimensions(in): 11.02×8.27×5.12 Cost ($): 990.00 Medical grade (y/n): no
	Verbal Analysis	- This syringe pump holds a single syringe; however, it can hold a variety of syringes, ranging from 10-60 mL. - It will alert the physician when: the injection is soon to finish, when the injection is done, if an occlusion occurs, if the syringe is improperly installed, a wrong setting is input, the device is at low battery, or if the syringe is loose. - Powered through wall power with 4 hours of battery life. - It should be stored between -30 to 55 °C. - It can operate with a single hand. Utilizes an LCD display with three work modes: rate mode, time volume mode, and dosage weight mode. - It is modular allowing for the stacking of these devices. This means that multiple chemicals could be injected at the same rate into the same subject.

Product Name (all names linked)	Description	
 **Alaris Syringe Pump Specs**	**Quantitative Analysis**	Accuracy: $\pm 5\%\ \dot{V} \geq 1\ \frac{mL}{hr}$ -8% to $+5\%\ \ \dot{V} < 1\ \frac{mL}{hr}$ Weight(lb): ~2.5 Dimensions (in): 3.3 x 8.9 x 5.5 Cost ($): Seems to be around $1000+ Medical grade (y/n): yes
	Verbal Analysis	- Capable of being a single, double, triple, or quad pump. Can attach four different modules at once. - Can administer large volumes of different kinds of fluids, IVs, blood, liquids, medication to the patient. - Can program to administer the exact amount of fluid. - Easily attachable to an IV pole - Maximum flow rate of 999 mL/h and minimum flow rate of 0.1 mL/h
 University of North Carolina (Greensboro) Research Pump	**Quantitative Analysis**	Accuracy: not posted Weight(lb): not posted Dimensions (in): 3.4 x 8.2 x 2 Cost ($): n/a open source Medical grade (y/n): no
	Verbal Analysis	- The pump was designed as a research project performed by Croatt Research group at the University of North Carolina. - The pump is largely 3D printed and designed to be open source. It uses motors like the NEMA 17 and readily available components. - The overall features are minimalistic; they have no way to detect occlusion.

Product Name (all names linked)	Description	
 Michigan Technological University Research Pump	**Quantitative Analysis**	Accuracy: ±1%(NEMA 11), ±5%(NEMA 17) Weight(lb.): ~2.5 Cost ($): n/a open source Medical grade (y/n): no
	Verbal Analysis	- This pump was designed as a research project by the department of materials science and engineering at Michigan Technological University. - The pump utilizes a lead screw method for actuation - They justified that both a NEMA 17 and 11 motor would not cause failure due to fatigue on the device. - This is for lab use and will not be used on patients
 Low-cost push-pull syringe pump	**Quantitative Analysis**	Accuracy: not posted Cost ($): 74.3 (component cost) - open source Medical grade (y/n): no
	Verbal Analysis	- This pump was designed as a research project meaning the end user of this pump is not medical, but for research purposes instead. - Their primary goal was to reduce the cost of the device. For this reason, a lot of their parts are 'off the shelf' or 3D printed. - In their paper they outline their calibration and testing process, which we can use as an outline for ourselves.

In addition to these pumps, our client has already drafted an initial design for this syringe pump, shown in Figure 3.

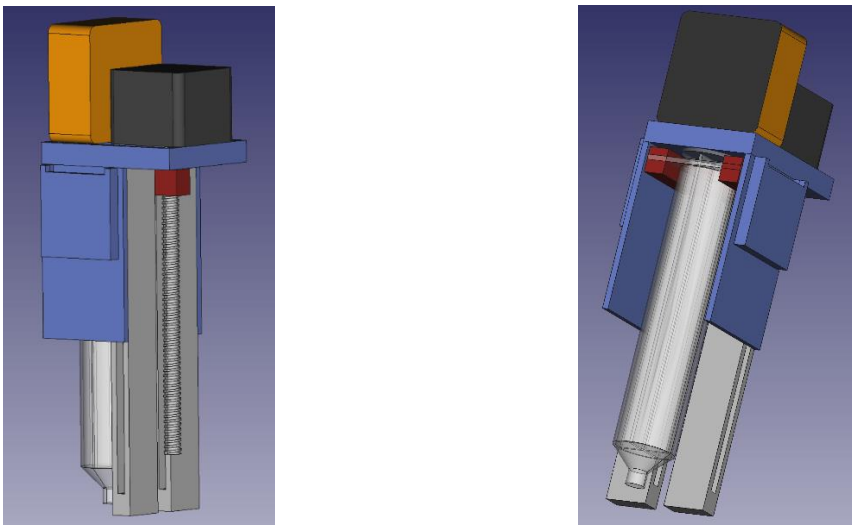


Figure 3: CAD model of client's solution

This provides us with a starting point. We analyzed the design and discussed the aspects of this design that stand out.

- Design utilizes simple shape geometry.
    - Since these parts are 3D printed, they will likely be quite fragile. The hard corners and edges also leave crevices that could be hard to sanitize. On the other hand, this will likely make the 3D printing process simpler.
- The lead screw actuation method is a compact way to get very accurate measurements.
    - However, the current method of constraint will likely either cause large frictional losses or have too loose tolerances.
- Design heavily relies on 3D printed parts.
    - This design does a great job of utilizing almost exclusively 3D printed parts and will be very cheap to produce because of it.
- Design pushes the syringe into the plunger.
    - This is a unique method of actuation. Our team was undecided on the benefits or drawbacks of this strategy. A potential downside is that this method moves more weight, which will cause unnecessary current draw on the motor.

With these aspects identified, we determined that there was enough that could be improved that it would be worthwhile to start from the ground up and take inspiration from the components of this design that stood out as positives.

Technical Specifications:

The project needs identified above are translated into technical specifications which give in-depth detail of quantitative values and rationale of the decision making towards a successful design of the syringe pump device. The list of technical specifications that were considered are attached below in **Error! Reference source not found.**.

Table 2: Technical Specifications

Project Need	Technical Specification	Rationale	Spec Evaluation
Volumetric Flow rate (1)	Device should be able to sustain a flowrate of 0 to 1200 $\frac{\text{mL}}{\text{hr}}$. (a)	Based on an interview with an anesthesiologist, this allows for the device to be used with a wide variety of anesthetics.	Performing flow rate test outlined later in the report.
	Flowrate should be indexable by 0.1 $\frac{mL}{hr}$ for flow rates less than 100 $\frac{mL}{hr}$ and 1 $\frac{mL}{hr}$ for flow rates above 100 $\frac{mL}{hr}$ (b)	Based on indexability of the industry standard BD Alaris Pump.	Performing flow rate test outlined later in the report.
	Accuracy of Volumetric flow rate should be less than 3% during majority of operation. (c)	Based on white paper published by Sendai Medical Center in Japan, they throw out all infusion pumps with an accuracy > 3% (link).	Performing flow rate test outlined later in the report.

Project Need	Technical Specification	Rationale	Spec Evaluation
Device Size (2)	The total mass of 1 syringe pump and the control module should be less than 2.5 lbs. (a)	Based on weight of market competitor, BD Alaris Syringe Pump.	Use Measure tool on CAD.
	The volume of the device should be less than 0.1 ft^3. (b)	Based on volume of market competitor, BD Alaris Syringe Pump.	Use Measure tool on CAD.
Hazard Prevention (3)	Device should be able to withstand 3 m drop without critical components fracturing. (a)	While the device is operating, it will be stored on an IV pole, therefore it should not be more than 3 m above the ground.	Perform Drop Test
	Device should alert user if pressure change is noticed. (b)	Outlined by IEC 60601-2-24 Ed1.0, Clause 51.5 the FDA's regulations which prevent hazardous outputs (link).	Performing occlusion test.
	Internal electronics should be protected from outside elements and have ventilation, excess liquid should be drained. (c)	Outlined by 60601 - 2 -24 Ed1.0, Clause 21.4 the FDA's regulations for syringe pumps (link).	Perform tests outlined in FDA document.
	Average temperature of microcontroller should not exceed $75°C$ (d)	Based on maximum temperature an Arduino operates at with $25°C$ buffer.	Measure internal temperature during operation.

Project Need	Technical Specification	Rationale	Spec Evaluation
Economic (4)	Total cost per device should be less than $100. (a)	Client provided this number.	Researching component costs and summing them.
	The total combined cost of prototyping should be below $4000. (b)	Based on funding proposal written by client.	Tallying all incurred costs and summing them.
Device Mounting (5)	When mounted the device should not slip. (a)	Our client requested this.	Performing slip test.
	Device(s) can accommodate 60 mL to 10 mL size syringes (b)	Most syringes used during operation are between 10 and 60 mL.	Test with varying sized syringe holder
Manufacturing (6)	Device should be manufactured using an Allen wrench, screwdriver, and 3D printer. (a)	The client believes these are the tools that hospitals will have access to.	Binary technical specification.
	The device should be made from 'off-the-shelf', and 3D printed parts. (b)	The client requested this.	Binary technical specification.

Project Need	Technical Specification	Rationale	Spec Evaluation
Sanitation (7)	All routine maintenance cleaning tasks should be able to be completed within 5 minutes. (a)	Based on the time it takes to clean BD alaris pump. (link)	Test with peers
	The device should use washable materials without a porous surface. (b)	Avoid sanitation fluid residuals trapped on the surface or damaging the cover material.	Binary technical specification.
Environmental (8)	All filaments that are used should be recyclable. (a)	Ensures the device does not produce needless waste.	Binary technical specification
Social (9)	The final product should be created using FreeCAD (a)	Requested by client: FreeCAD is easily accessible for organizations who do not want to pay licensing fees.	Binary technical specification
Syringe Interface (10)	Syringe should remain secured in apparatus when acted on by an axial force of 500 N and a transverse force of 100 N	Based on maximum anticipated axial and transverse loads.	Perform experiment outlined in Variable Syringe Mechanism Experiment
	Device should rigidly (defined in 10a) support syringes that vary in size from 10-60 mL	Specified by client	Perform experiment outlined in Variable Syringe Mechanism Experiment

Design To Date:

Current Design Overview:

The current design has four important components:

1. Structure
2. Parallel Syringe Pump Mechanism
3. Actuation Mechanism
4. Variable Syringe Mechanism

These four components accomplish the primary goals of this device effectively as shown in the technical specification evaluation later in this report. The current design, shown in the figure below, is still rather bulky. Moving forward, we are looking to optimize the design further by cutting out unnecessary structural components. However, the actuation method is working very well and producing ideal outputs, and both the parallel mounting mechanism and the variable syringe mechanism are functional.

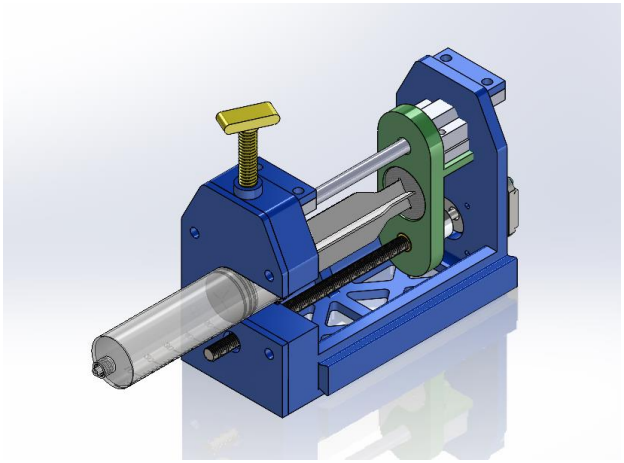


Figure 4: CAD model of design to date

A more in-depth analysis of each of these components follows.

Structure:

The entirety of the devices structure is currently 3D printed. It is one homogenous 3D printed component making it trivial to manufacture if the user has the requisite resources. The original structure design, shown below, was much bulkier than the current design. As each of these parts will ultimately be 3D printed, this bulk is directly correlated to print times which we are striving to reduce.

An FEA simulation was performed on the primary structure to identify high stress locations that are likely to fail. These regions will be systematically supported to further increase the durability of the structural body.

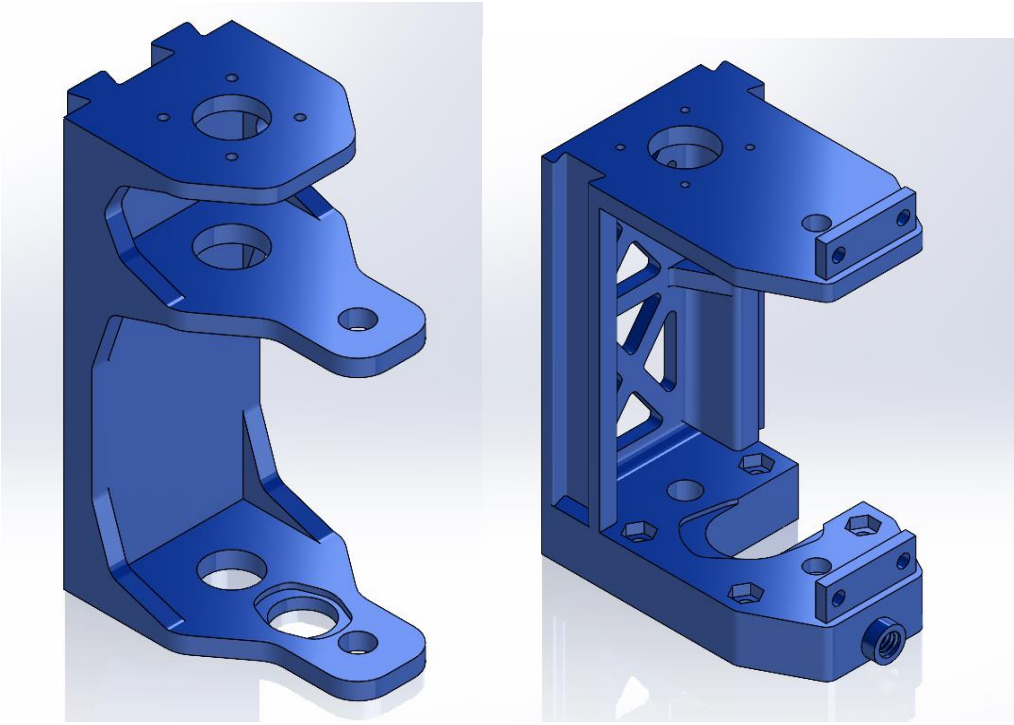


Figure 5: The original design for the structure (left) was bulky and did not allow for assorted syringes to be inserted. The new design (right) is smaller and has the capability to interface with a wide variety of syringes.

The volume reduction happened in two primary locations. The first location was merging the platform that the stepper rests on with the platform used to support the guide rail. This design decision greatly reduced the print time, and the only shortcoming is the lead screw is slightly too long and sticks out the end.

The other area of improvement was pocketing low stress areas. The bottom was pocketed based on an FEA simulation that indicated negligible stress in that region. While we considered pocketing the platform that the stepper rests on, this would be introduce additional supports in this region, so the time reduction was negligible. This leads to a slightly stronger product.

Table 3: Volume and Print Time comparison

Quantity	Quarter 1 value	Quarter 2 value	Percent difference
Volume (in^3)	19.8	16.8	- 17.8 %
Print Time (hr)	18.5	19.7	6.1 %

The volume has been slightly reduced of the structural component, next quarter the team will analyze reducing the print time. Ideally the volume changes directly correlate to a lower print time, however the print time actually increased. This is from the additional component added to the first prototype to account for different size syringes, as the design now requires more support material compared to the first prototype design.

To reduce this time the team will investigate the possibility of utilizing tree supports rather than the autogenerated supports that are currently being used. The team will continue to look for additional areas to reduce the volume of the structure.

Parallel Syringe Pump Mechanism:

To attach multiple pumps to the device in parallel, we are utilizing a single dovetail slot on one side and a slot on the other side of the syringe body. This will allow the user to easily slide the syringe pump bodies into a parallel position. As shown in Figure 5 below.

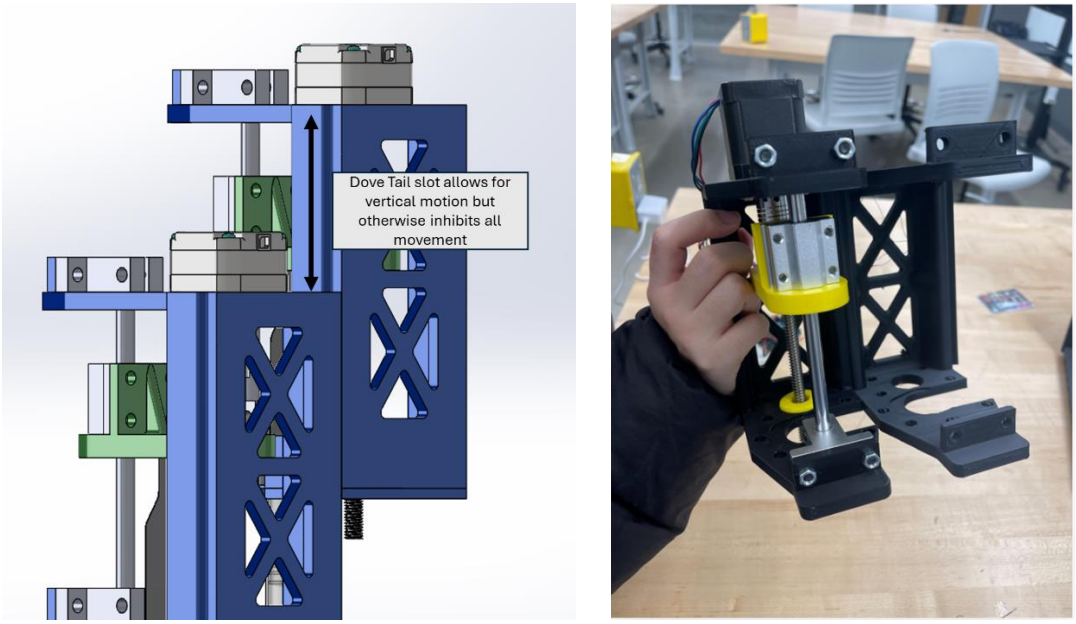


Figure 6: The parallel syringe pump mechanism allows for multiple pumps to be attached together with negligible deflection.

From testing this mechanism, a small amount of friction is produced between the material surfaces which makes it somewhat difficult to slide these in place. However, once the two bodies are slotted, the friction helps with keeping both pump bodies in place. Looking forward the team will try to reduce the friction by reducing the tolerances between these two parts and potentially introduce a simple locking mechanism.

The major success with this mechanism is its simplicity. It adds negligible time to the main body print and accomplishes the desired task effectively. While the locking mechanism will be

explored, if it adds to the complexity or print time, the current design may be the final version prototype design for this mechanism.

Actuation Mechanism:

The actuation mechanism was selected after a rigorous concept selection process that was the focus of the first quarter. This process is expanded upon in Appendices B and C. Essentially, the team selected the lead screw mechanism because of its simplicity, cost effectiveness, and consistency. Since a large amount of time was dedicated to modeling this mechanism, relatively few changes have been made to it.

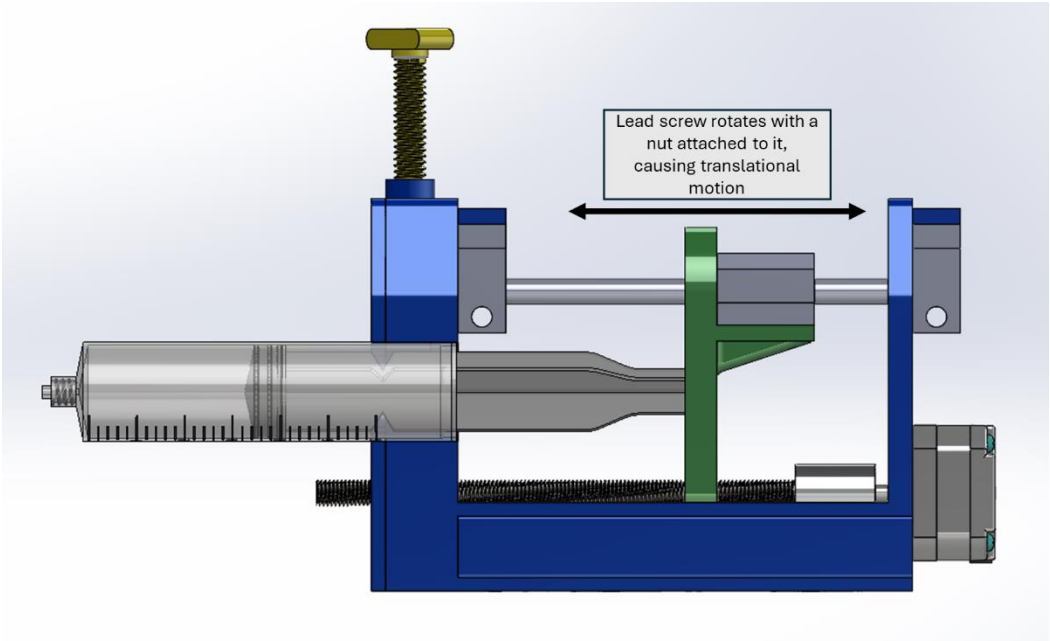


Figure 7: The proposed actuation strategy in quarter 1 was effective at meeting all technical specifications.

This quarter, the emphasis has been on testing the validity of this actuation device. It has been determined that the lead screw method is effective at hitting both the range and indexability technical specifications, but without closed loop control it is not as accurate as was initially anticipated.

Variable Syringe Mechanism:

From looking at technical specifications, a variable sized syringe mechanism was constructed to provide additional support for syringes that are smaller than 60 mL in volume. The prototype of the design consists of two 3d printed casing halves and a gripper. The gripper consists of a 3D printed lead screw and gripper block at the end, placed within the two casing halves that are held together with screws shown in Figure 5.

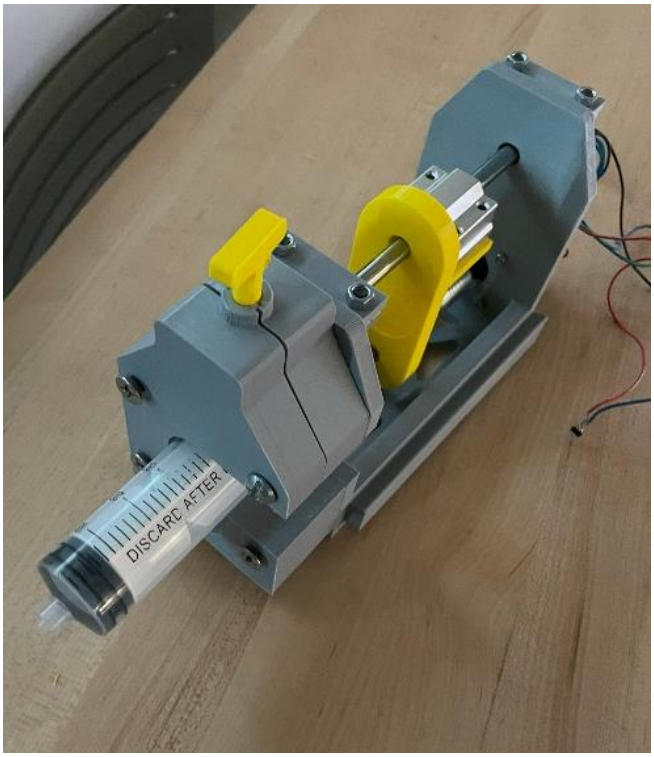


Figure 8: Initial Variable Syringe Mechanism Design Attached to Syringe Pump Body

After further discussion, the bulk of the design was a concern. The improved design of the variable sized syringe mechanism is implemented to the base plate of the syringe pump body. The size height of the case was greatly reduced, and the lead screw and gripper block system were maintained. The inner thickness of the edges was also increased to increase the stability.

Figure 9: Updated variable syringe mechanism design attached to syringe pump body.

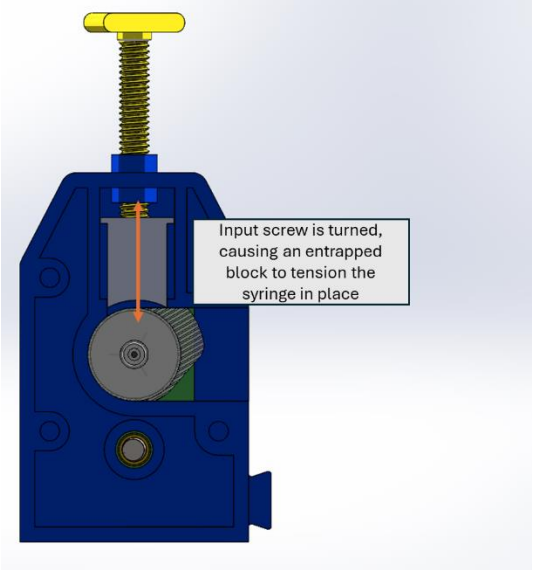


Figure 10: Lead screw gripper system

The figure above shows a cross sectional view for the variable syringe mechanism. This system utilizes two small 3d printed components: an entrapped slider and a screw. At the moment, the team is experimenting with swapping the 3d printed screw out for a metal screw to further reduce print and assembly times. This would also provide a longer lifespan for this mechanism.

Design Validation:

Finite Element Analysis was conducted on body base design. Within the software, a force of 0.875 lbf is pointing downwards from the lip edge on where the syringe is resting. This force simulates the stepper pushing the plunger of the syringe to displace the liquid. Another force is appointed at the slot where other syringe bodies would be attached. Additionally, a bearing load has been applied on the smaller holes on the base to simulate the force from the rod support.

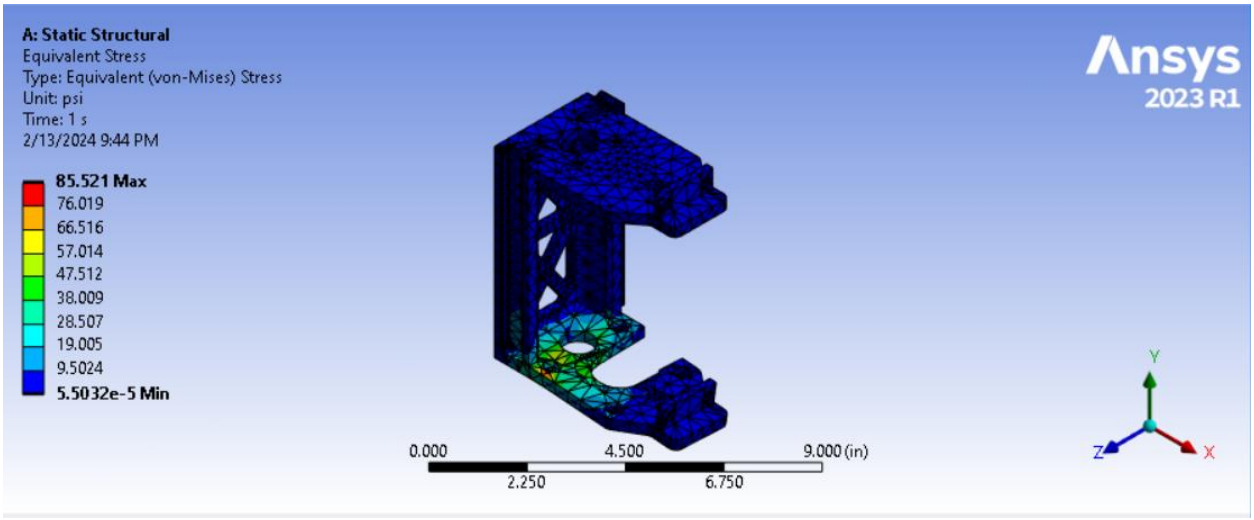


Figure 11: Total Equivalent Stress on Syringe Pump Body

The Finite Element Analysis on the syringe pump body for total equivalent stress shows that the body is capable of withstanding the force that the stepper will be outputting onto the syringe. From Figure the amount of stress from the stepper acting on the syringe is minimal.

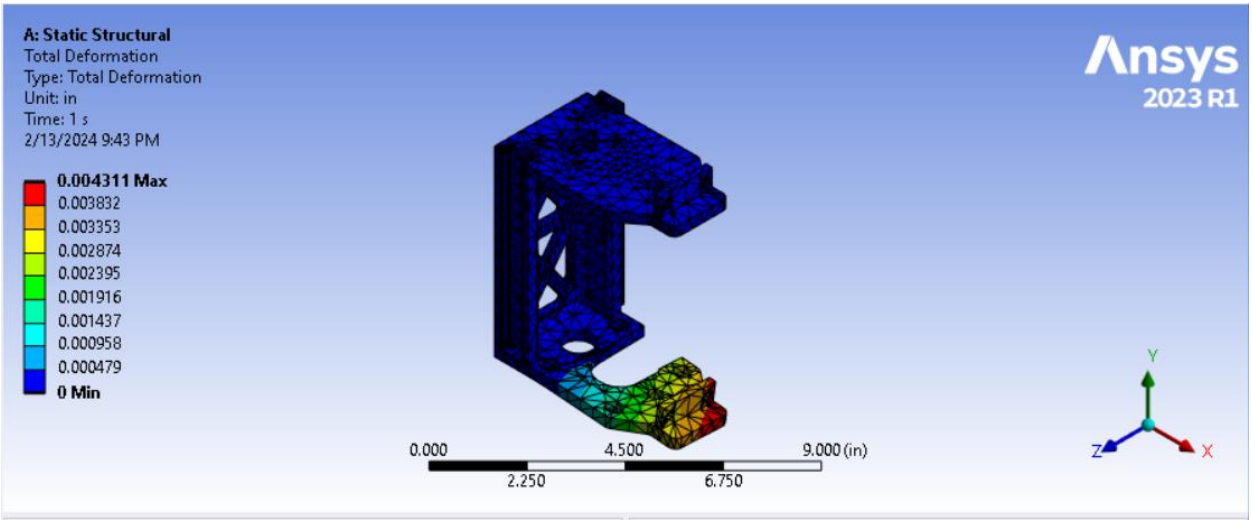


Figure 12: Total Deflection Analysis on Syringe Pump Body

Additionally, based on the Finite Element Analysis of the body, there is high stress and deformation at the extending edge of the base plate. Having the variable sized syringe mechanism combine to the base would add additional thickness to base plate which will hopefully provide additional support at the high stress concentration area.

Technical Specification Evaluation:

Table 4: Technical Specification Evaluation

Project Need	Technical Specification	Evaluation Method	Status
Volumetric Flow rate (1)	Device should be able to sustain a flowrate of 0 to 1200 $\frac{\text{mL}}{\text{hr}}$. (a)	Flowrate experiment explained in Preliminary testing of Prototype section.	Met. Device is capable of sustaining flowrates below 4500 $\frac{mL}{hr}$.
	Flowrate should be indexable by 0.1 $\frac{mL}{hr}$ for flow rates less than 100 $\frac{mL}{hr}$ and 1 $\frac{mL}{hr}$ for flow rates above 100 $\frac{mL}{hr}$ (b)	Flowrate experiment explained in Preliminary testing of Prototype section.	In progress. Device is capable of indexing at 1 $\frac{mL}{hr}$ at flow rates above 100 $\frac{mL}{hr}$, but low flow rate indexability has not been tested
	Accuracy of Volumetric flow rate should be less than 3% during majority of operation. (c)	Flowrate experiment explained in Preliminary testing of Prototype section.	Not met, will require a feedback controller.

Project Need	Technical Specification	Evaluation Method	Status
Device Size (2)	The total mass of 1 syringe pump and the control module should be less than 2.5 lbs. (a)	Weigh prototype on scale	Met. The current prototype weight is 0.25 lb.
	The volume of the device should be less than 170 in^3. (b)	Use Measure tool on CAD.	Met. The Current size is 16.7 in^3.
Hazard Prevention (3)	Device should be able to withstand 3 m drop without critical components fracturing. (a)	Perform Drop Test	Not tested.
	Device should alert user if pressure change is noticed. (b)	Performing occlusion test.	In progress. In collaboration with ECE team and initial tests have been performed.
	Internal electronics should be protected from outside elements and have ventilation, excess liquid should be drained. (c)	Perform tests outlined in document.	Not tested.
	Average temperature of microcontroller should not exceed $75°C$ (d)	Measure internal temperature during operation.	Not tested.

Project Need	Technical Specification	Evaluation Method	Status
Economic (4)	Total cost per device should be less than $100. (a)	Researching component costs and summing them.	Not Met. Current cost is: $141.76.
	The total combined cost of prototyping should be below $4000. (b)	Tallying all incurred costs and summing them.	Met. Current cost is: $1246.60
Device Mounting (5)	When mounted the device should not slip. (a)	Performing slip test.	Not Tested.
Manufacturing (6)	Device should be manufactured using an Allen wrench, screwdriver, and 3D printer. (a)	Binary technical specification.	Met. Assembly requires Phillips head screwdriver, Allen keys, and 3D printer.
	The device should be made from 'off-the-shelf', and 3D printed parts. (b)	Binary technical specification.	Met. All parts are either purchased on amazon or 3D printed.

Project Need	Technical Specification	Evaluation Method	Status
Sanitation (7)	All routine maintenance cleaning tasks should be able to be completed within 5 minutes. (a)	Test with peers.	Not Tested.
	The device should use washable materials without a porous surface. (b)	Binary technical specification.	Met. Device is made from aluminum and pla (petg for final prototype)
Environmental (8)	All filaments that are used should be recyclable. (a)	Binary technical specification	Met. All filament is recyclable.
Social (9)	The final product should be created using FreeCAD (a)	Binary technical specification	Not Met. Current prototypes have been made in SolidWorks for faster turnaround times. Will be converted at the end
Syringe Interface (10)	Syringe should remain secured in apparatus when acted on by an axial force of 10 N and a transverse force of 5 N	Perform variable syringe mechanism experiment.	Met, withstood >30 N in the axial direction and >10 N in the transverse direction.
	Device should rigidly (defined in 10a) support syringes that vary in size from 10-60 mL	Perform variable syringe mechanism experiment.	Met. All syringes fit within the apparatus.

Preliminary Testing of Functioning Prototype

Flow Rate Experiment:

The focus of the past quarter has been on the linear actuation of the syringe pump. To provide a final verification that selected concept would provide the desired accuracy, an experiment was designed to test the output of our actuation method.

Motivation:

- Identify any deviation in the flow rate from the expected.
- Identify the range of flow rates that the actuation concept can support.
- Compare results to mathematical model.

Equipment:

1. Syringe pump.
2. VWR scale.

Experimental Apparatus:

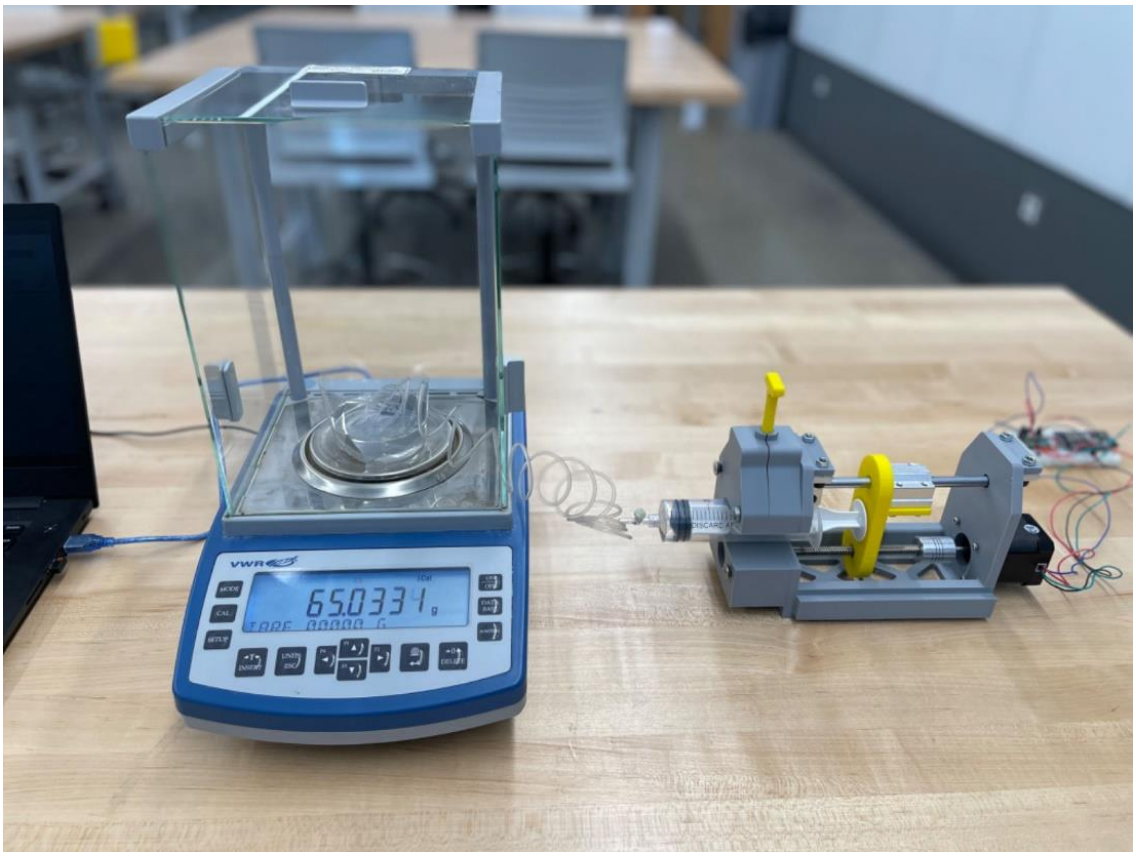


Figure 13: Experimental Design for Volumetric Flowrate Test

Procedure:

1) The device is setup as shown in Figure 12, the syringe is filled with water.
2) The VWR scale is connected to a computer to record the data.
3) A delay time is set in the Arduino controller to adjust the rotational speed of the motor. The delay time ranges from 200 to 5000 microseconds. As the delay time increases, the rotational speed of the motor decreases.
4) For each delay time (which corresponds to a rotational speed), the total mass of the pumped water in the cup on the VWR scale with a uniform time step (0.5s) is recorded.
5) The process from 4) is repeated twice for each motor speed.

Sample Data:

In step 4, the digital scale records the accumulated mass of the water pumping into a cup. The figure below captures the general increasing trend of the total mass of the water. The data could be interpreted as 3 segments: the preparation region where the mass is slightly increased and the rotational velocity is increasing from zero, the linear region where the mass is linearly increased and the rotational velocity has already achieved the target value, and the ending region where the mass stops increasing after the motor is stopped.

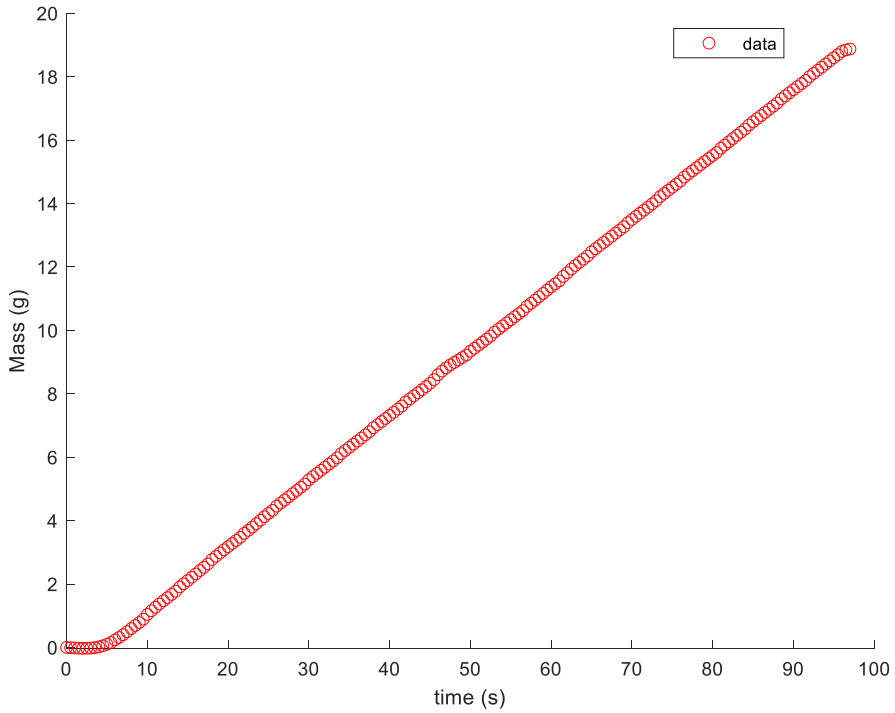


Figure 14: Raw Data of the Accumulated Mass over Time for Delay Time at 3000 Microseconds

Data Processing:

For each delay time, we take the average flowrate from two test trials as the final flowrate value. In the end, we can draw this following figure aiming to find a relationship between the flowrate and the delay time. We use an inverse power curve to calibrate the flowrate for different delay times based on the mathematical model.

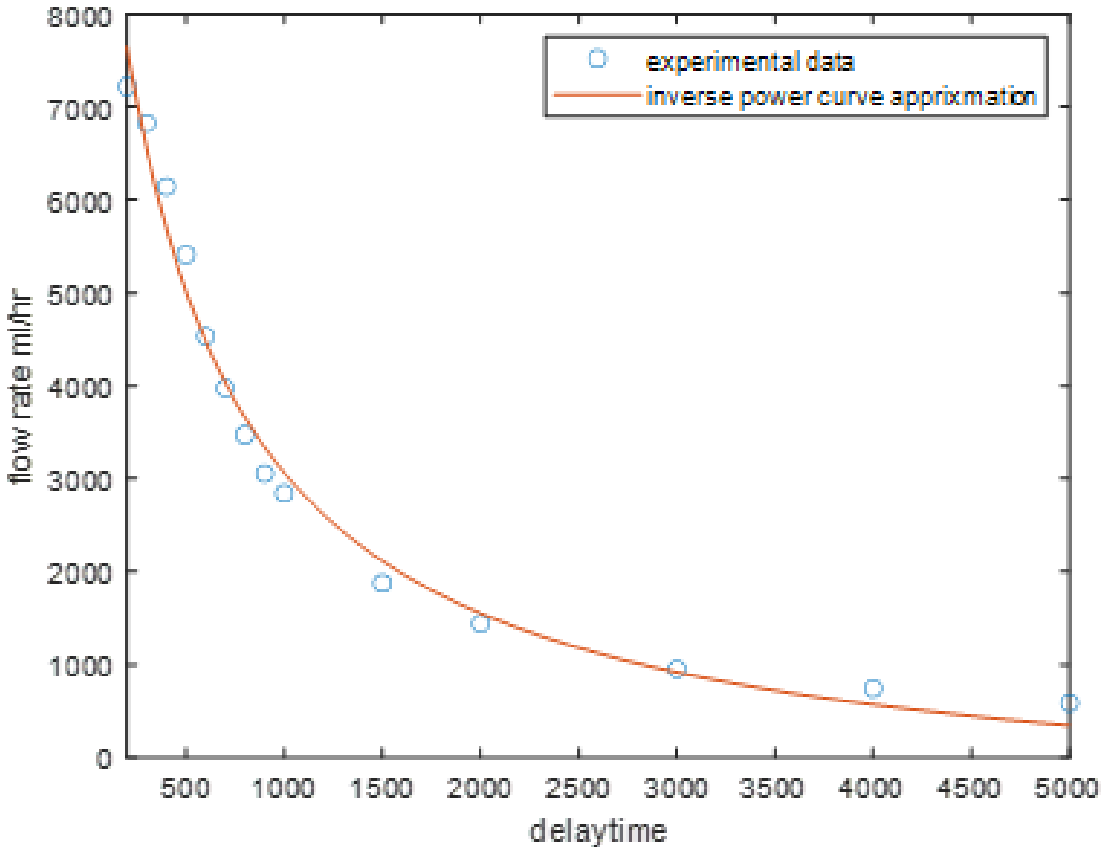


Figure 15: Average Flowrate for Each Delay Time

From the raw data, we use the 2nd order accuracy finite difference method to calculate the volumetric flowrate at each time step. In the figure below, the flowrate increases dramatically in the preparation region. After about 5 seconds, the motor achieves the desired flowrate, and instantly the flowrate fluctuates on the target flowrate (at the red horizontal line). This transient response is much smaller than the total time it takes to empty the syringe.

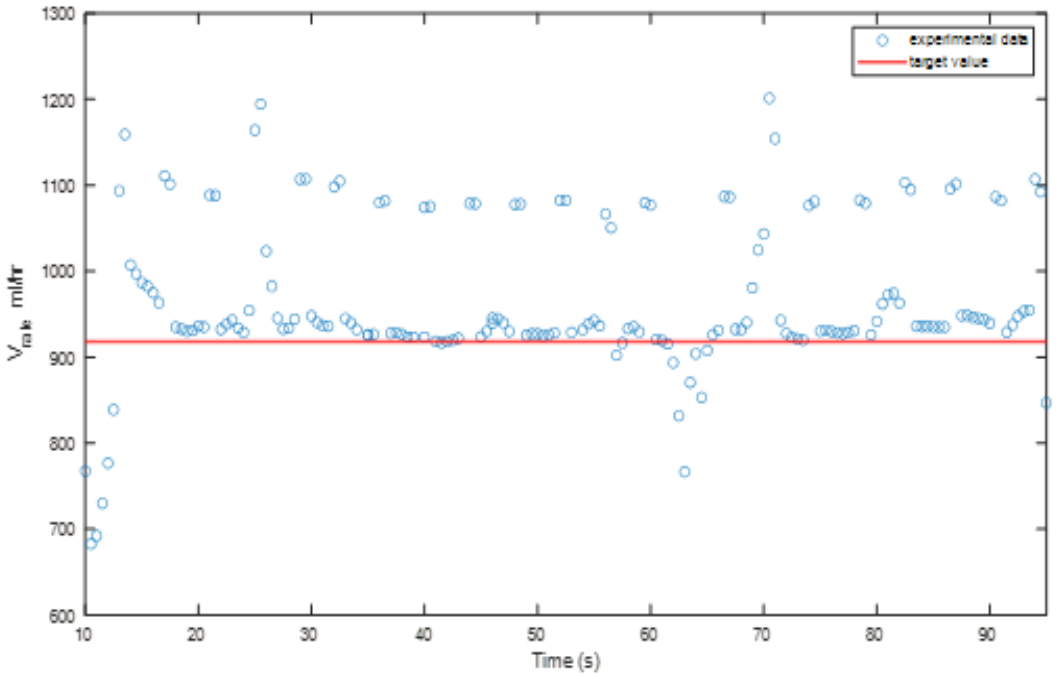


Figure 16: 2nd Order Approximation of Volumetric Flowrate for Delay Time at 3000 Microseconds.

Future Work:

The average flowrate for each delay time resembles the multiplicative inverse function of what we expected from the mathematical model. In the higher flowrate region, the stepper motor would lose a step if the speed was approaching the maximum which is why the experimental speed is below the theoretical value for a small-time delay. In the future, we need to perform the flowrate experiment again with the new close loop controller produced by the ECE team.

Back Pressure Experiment:

Motivation:

We need to select a stepper motor that is strong enough to push the fluid within the syringe to simulate the impact of a patient's blood pressure.

Equipment:

1. Syringe pump
2. Medicinal tubing

Diagram:

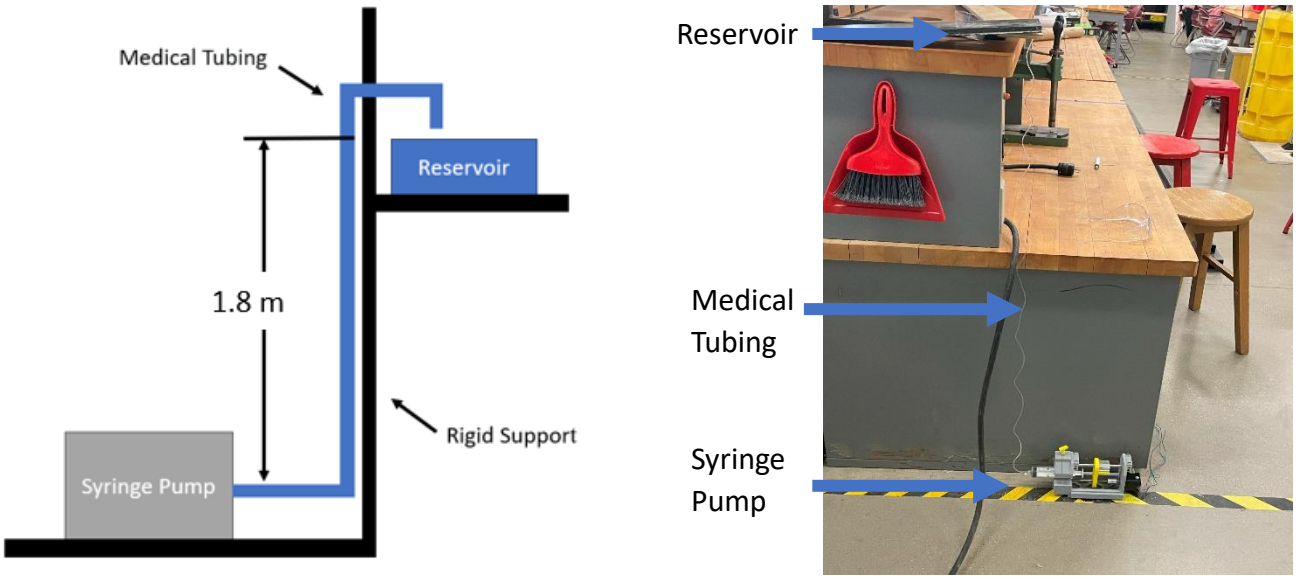


Figure 17: Back Pressure Experiment Design for Blood Pressure

Procedure:

1) A 60 mL syringe is loaded with 60 mL of water.
2) The tubing is attached to the end of the syringe and the output end is put to a height of 1.8 m from output of the syringe
3) The time delay between each step is set to 200 μs (our minimum delay time) in Arduino code for the first trial.
4) The syringe is then reloaded and retested at a flowrate of 2000 μs (our maximum delay time).
5) The experiment is repeated with a different size syringe.

Summary:

For this experiment, the team decided to just observe if there is fluid flowing out of the tubing after traveling to the specified height. From the diagram above, the height was determined to be

1.8 meters from the base of the syringe where the fluid is being displaced into the tubing. Due to 1.7 m of tubing that the team had access to, we were only able to verify a height of 1.7 m. The height value was obtained from utilizing the definition of hydrostatic pressure.

$$P = \rho g h$$

Where, P is the high blood pressure, ρ is the density of blood, g is the gravity and h is the height. From outside sources, $P = 140\ mm\ hg = 18665\ \frac{N}{m^2}$ for high blood pressure, $\rho = 1050\ \frac{kg}{m^2}$ and $g = 9.81\ \frac{m}{s^2}$. In solving for the height, the value obtained is 1.8 meters.

After completion of the experiment, the stepper was able to push the fluid to a height of 1.7 meters for both delay times of 200 μs and 2000 μs with the larger 60 mL syringe. Using a smaller sized syringe resulted with similar results. A conclusion can be made that the stepper will be able to overcome the back pressure regardless of different syringe sizes and delay times within the 200 μs and 2000 μs range.

Variable Syringe Apparatus Experiment:

The final stretch of this quarter has been focused on providing a method for syringes of various sizes to interface with this device. This experiment quantifies how effective our mechanism is.

Motivation:

- Identify the maximum axial and transverse forces the variable syringe mechanism can support.
- Identify the effect syringe size has on this relationship.

Equipment:

1. Syringe Pump
2. Digital Force Guage

Experimental Apparatus:

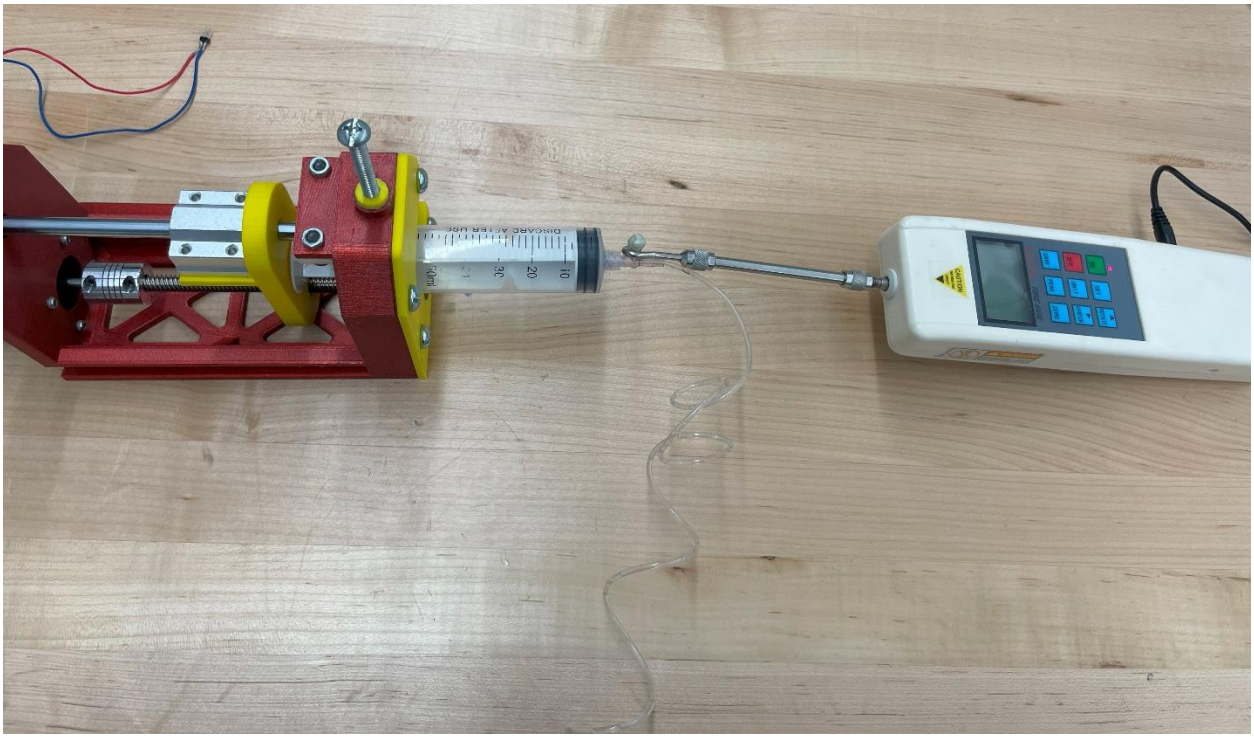


Figure 18: Experimental design for the variable syringe apparatus test.

Procedure:

1. Attach the digital force gauge (with the hook attachment selected) to the syringe output.
2. Clamp the syringe pump down to ensure that it does not move.
3. Pull the force gauge directly away from the pump and record the peak force.
    a. If the force gauge reaches 30 N stop pulling
    b. If the syringe slips out stop pulling

4. Pull the force gauge perpendicularly away from the pump and record the peak force
    a. If the force gauge maxes out stop pulling
    b. If the syringe slips out stop pulling
5. Repeat steps 3-4 10 times to ensure consistent results.

Summary:

The above procedure will be run for each of the syringes we have access to. The trends can then be analyzed. We selected the two directions of the force based on the technical specifications defined in the section above. The perpendicular force is exerted since that is the most likely direction that the syringe slips out in, however it somewhat unlikely to occur, and would only be caused by an external factor. The more likely 'axial' force will be constantly exerted by the stepper on the syringe plunger. The structure of the pump is designed to reduce this effect, but this force will in turn be exerted onto the variable syringe mechanism.

It was determined that we would cap the force exerted at 30 N. This should be fine since the technical specification just requires a axial and transverse force of 5 N to be considered met. In doing so, we prevented any unnecessary damage to the device and ensured we would not cause failure in any of the other components.

Sample Data:

Table 5: The variable syringe test indicated that the mechanism was capable of withstanding both an axial and transverse force. However, the mechanism is not as well equipped to handle a transverse force.

	60 mL Syringe		**10 mL Syring**	
Trial #	**Axial Force**	**Transverse Force**	**Axial Force**	**Transverse Force**
1	Max	14.2	Max	10.0
2	Max	12.9	Max	9.6
3	Max	13.7	Max	11.3
4	Max	13.8	Max	10.8
5	Max	14.3	Max	9.7
6	Max	17.2	Max	10.2
7	Max	16.3	Max	10.5
8	Max	17.7	Max	11.2
9	Max	16.6	Max	9.7
10	Max	15.4	Max	11.7
Average	**-**	**15.2**	**-**	**10.5**

Data Processing:

At both the largest and smallest syringe size, the technical specification is met. This is ideal, however, based on these two data sets there is a downward trend in the transverse force that the system can withstand.

Future Work

At this moment, the team only has access to 60 mL and 10 mL syringes. These syringes are the upper and lower bounds of the syringes the system could interface with, so we anticipate that most other syringes within this range will also fall in range assuming they maintain a simillar geometry to the ones we tested (generally cylindrical shape with a thumb rest). Despite this, it would likely be ideal to test with a wider range of syringes.

Plans for next Quarter

Next quarter will begin with a focus on refining the design based on feedback and work on implementing the controller from the ECE team into the design. The 3D printer has been ordered so we are hoping to experiment with different infill levels and test using PETG instead of PLA. The print times for our design also need to be monitored and see if they can be reduced using different printing modes. In addition, there are a few tests that still need to be completed to address some of our technical specifications pertaining to our final design.

Figure 19: Tentative Gantt Chart for Quarter 3

The majority of the second half of the final quarter will be focused on wrapping up this project. Most of the CAD work to this point has been performed in solidworks. This made the prototype phase progress very rapidly, however, is not as accessible as other software. After a brief discussion with the team, we are tentatively moving forward to replicate the model in freeCAD, though we are also potentially interested in working with OnShape. A user manual will also be compiled to ensure that the work we have done is easily accessible to those who want replicate to the design.

Acknowledgements:

The team would like to thank Drs. Bernal and Sangelkar for their continued support throughout this project. Their weekly meetings have kept the team on track and ensured our success. Their feedback has largely guided our progress, by taking the time to ask probing questions they have not only let us develop as engineers, but also let us deliver a successful product.

The insight from Dr. Dugan from the external review has had an immeasurably positive impact on the team. Her insight and experience have greatly helped us scope our project to best serve the end user.

The ECE team (Jake Armstrong, Tyra Correia, Gabe Neise, and Wei Huang) has also directly contributed to our success. All our experimental trials were directly reliant upon the ECE team to produce a functional prototype very early. Their flexibility and willingness to help has let our team achieve much more meaningful results.

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Appendix A – Ethical Considerations

The goal is to design a medical device that can be used for people in developing countries. To ensure this device is beneficial, the team has identified the following considerations and explained how we strive to address these concerns with our design process. These considerations appear within our technical specifications as well.

Table A1: Ethical Considerations for this Project

	Considerations for the project
Public health, safety, and welfare	• Ensure all models we utilize are accurate, and if they are not accurate, report the uncertainty and any regions where the behavior of the system does not behave as anticipated. (Technical specification 1c) • Build redundancy into the system, particularly with regards to emergency stops. • Test the device in 'catastrophic' circumstances (e.g., water spills on screen, pump falls of pole). Ensure device can continue standard operation while this happens, or that it can quickly be stopped. (Technical specifications 3a, 3c)
Cultural	• Ensure all documentation we provide is written in simple English. A doctor with a weaker grasp on English should still be able to read the user manual and understand the intent of each statement.
Social	• All the work is open source. (Technical specification 9a)
Environmental	• The device is meant to be reusable (minimum 1 year life cycle), if used with care and periodic cleaning. (Technical specification 8a) • Considering the potentially massive amount of production, the 3D print material should be recyclable. (Technical specification 8b)
Economic	• The device is made for developing and underdeveloped countries that lack medical equipment and funds. o Therefore, all components should be priced reasonably (Technical specification 4a)

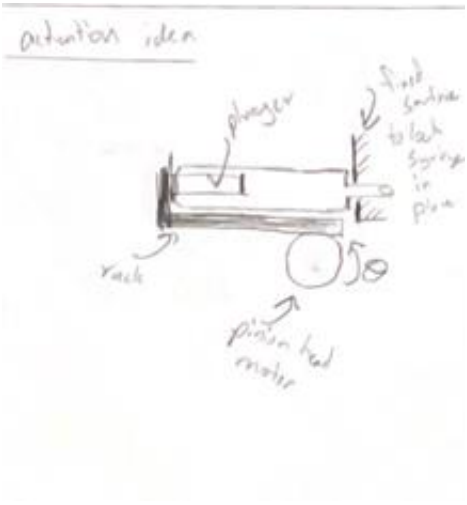
	Considerations for the project
Sustainability (End of Life Cycle)	• Casing and other non-critical components should be made with material that is recyclable or easily reusable. (Technical Specification 8b) • Components that touch chemicals (e.g., syringe, plunger) should be made so that they can easily be cleaned prior to disposal in order to reduce chemical waste (or they should be treated as medical waste). (Technical Specification 7c)
Engineering Standards	• It should meet all the FDA's regulations outlined in IEC 60601-2-24 which applies to all infusion pumps. Getting past this regulation is a very steep requirement, this should also ensure that this device is not a threat to public health or the user's safety. (Technical Specification 7a, 7b, and 7c)

Appendix B – Concept Generation:

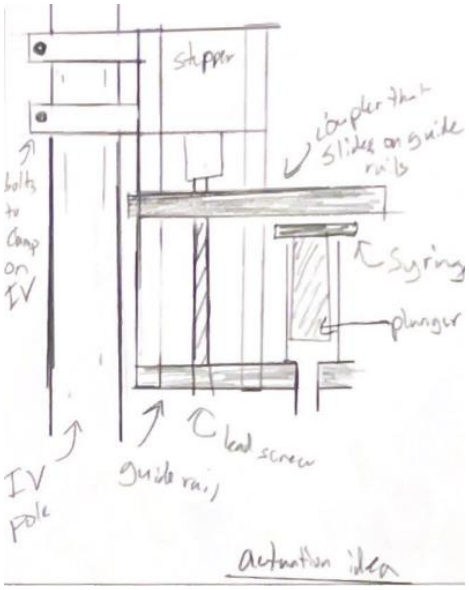
The concept generation process we utilized was based on the collaborative sketching model. We first identified three vital functions: an actuation method, securing variable sized syringes, and mounting to an IV pole. Our team of three each drew up four concepts for each of these functions within about 20 minutes. In this time, we discussed together and researched to see how these functions are typically achieved. These sketches were passed to the next person, and they added ideas to the original concepts, this was then repeated once more to get a slightly more flushed out concept. All our ideas from this process are shown in Appendix A.

The concepts that we focused on are shown alongside an initial list of pros and cons that we drafted for each concept.

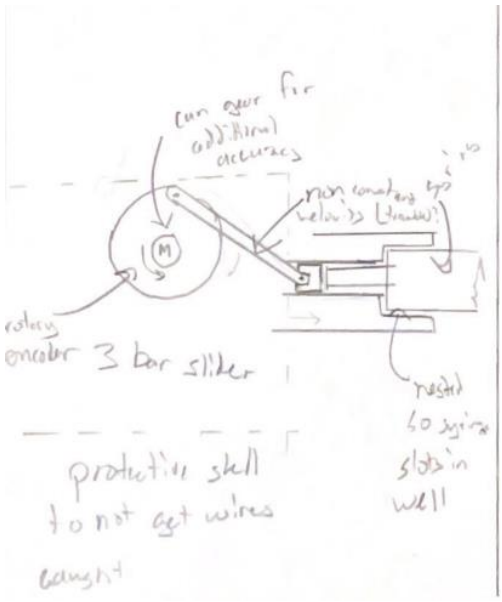
Description:	**Rack and pinion to translate rotational motion to linear motion**
Pros:	• Simple Design • Easy to mathematically model
Cons:	• Rotary Encoder needed • Bulkier than alternatives • Challenges with wiring



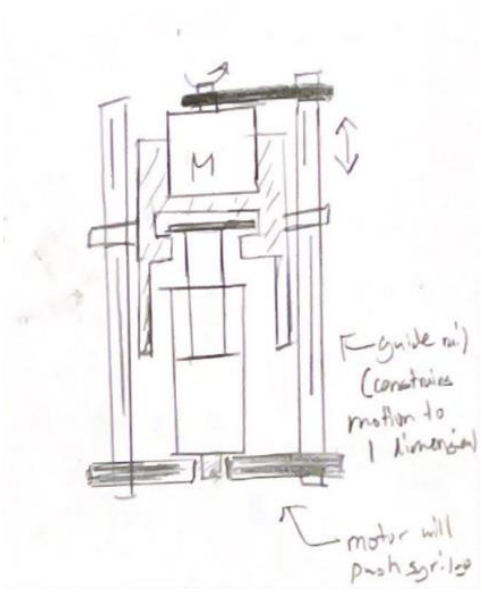
Description:	Stepper motor – lead screw linear actuation
Pros:	• Accurate • Less wide • Easy to fasten to IV pole
Cons:	• Exposed lead screw • Fragility • Hard to clean • Data Processing with Stepper motor having a non-constant angular velocity can be a challenge



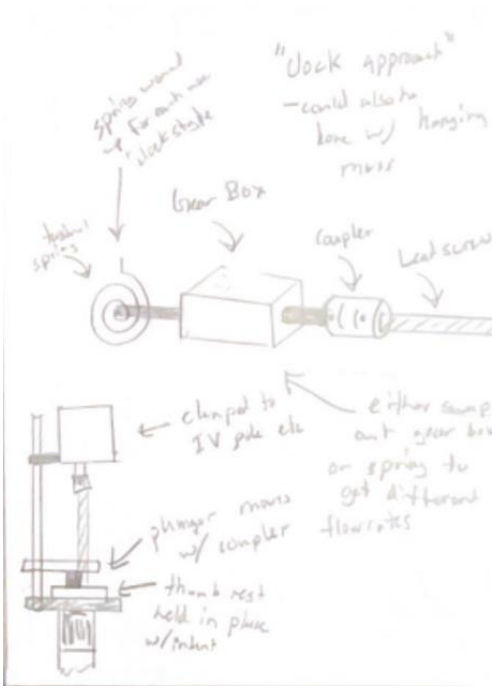
Description:	Piston Cylinder
Pros:	• Accurate • Cheap (only uses DC motor) • Easy to load and unload syringe
Cons:	• Require PD control to approximate constant velocity. • Bulky • Need rotary encoder



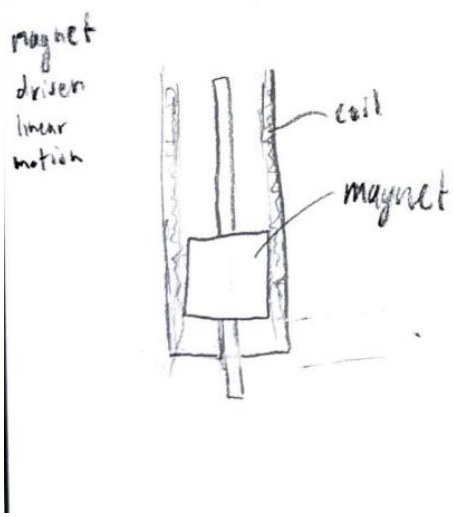
Description:	**Backward motion actuator**
Pros:	• might be easy to hold the syringe from the needle part.
Cons:	• Infusion tube might be pulled out from the patients. • The size is larger than the backwards pushing way



Description:	**Clockwork mechanism**
Pros:	• Would not require ECE components. • Very reliable without access to power
Cons:	• Lots of complex parts • Hard to mathematically model • May be inaccurate



Description:	**Electromagnet coil**
Pros:	• Does not take up much space. • Easy setup • Does not need an external motor, since the whole system is a big "motor." • Might consume more electricity
Cons:	• Need protection on the coil for excess fluid. • Relies on electromagnetic interaction and advanced physics concepts. • Would require a large amount of power. • Would require a major pivot for the ECE team.



Appendix C – Concept Selection:

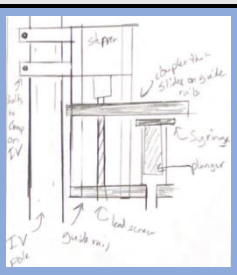
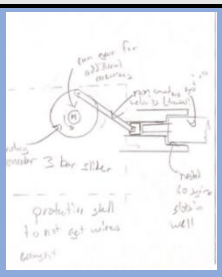
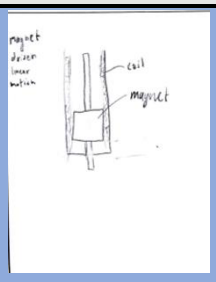
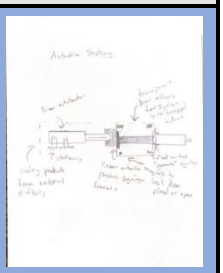
To identify the optimal solution from the concepts we generated, we first went through a discussion phase where we threw out any ideas that we deemed unrealistic. Ideas like the clockwork mechanism were considered unrealistic because of their demand for precise parts, whereas ideas like the linear slides were thrown out because it would be too bulky.

After this, we designed a Pugh matrix that ranked each of our valid concepts. We ranked them based on the criteria defined below:

> **Low Cost:** The combined ME and ECE cost of the device should be less than $100. This means that we are heavily constrained with the components that we use, and concepts that are reliant on things like stepper motors or linear actuators are going to be penalized in this section.
>
> **Low Bulk:** The device should be light and compact, solutions that rely upon large components or have parts that require steel or other heavy material will be penalized.
>
> **High Simplicity:** Since the device is to be built by untrained workers, it should be able to be made with exclusively a 3D printer and simple tools (screwdriver, wrench, etc.). Concepts that will rely on more complex manufacturing strategies will score lower in this section.
>
> **High Durability:** The environment that the pump will be operating in will not be ideal. This means that the device should be able to withstand regular transport and drops from IV poles. This will largely penalize designs that are overly complex or have a lot of moving parts.
>
> **High Reliability:** We defined reliability to be how consistently the device works, and how prone to jamming or sudden failure the device is. This is going to penalize designs that have lots of moving parts or rely upon less consistent mechanisms. This category will also penalize concepts that will not have a constant linear output velocity or will require a controller to produce consistent outputs.
>
> **High Accuracy:** Since this is a medical device, the accuracy of the flow rate coming out of it is very important. The accuracy of the flow rate out will be directly correlated to the positional accuracy of the actuator. This category will penalize all devices that will produce an output that is not accurate or that cannot be easily measured.

We ranked all our valid solutions using the Pugh matrix found in Table .

Table C1: Pugh Matrix for Actuation Concept Selection

	Concepts				
Selection Criteria	 **Stepper Motor / Lead Screw**	 **Piston Cylinder**	 **Rack and pinion**	 **Magnetic Coil**	 **Linear Actuator**
Low Cost	-	-	Datum	+	-
Low Bulk	-	-		+	-
High Simplicity	0	-		-	+
High Durability	+	-		0	+
High Reliability	+	-		-	+
High Accuracy	+	+		-	+
Sum +	3	1	Datum	2	4
Sum -	2	5		3	2
Sum 0	1	0		1	0
Net Score	1	-4	0	-1	2
Rank	2nd	5th	3rd	4th	1st

From this table, we have selected five designs from our concept generation which we thought were the front runner concepts. We selected the Rack and Pinion design to be the datum as we saw the concept to be moderate in terms of difficulty to design and function. In addition, team members had experience working with rack and pinion mechanisms, this meant that accurately quantifying this was easier than other concepts. It is easier for us to compare these concepts to something that we already understand rather than comparing it to a mechanism we have never worked with before.

However, we wanted to further thin the field prior to creating a weighted decision matrix and doing a deep dive into each topic. We decided to move forward with the top three ideas identified with our pugh matrix. While we did look at continuing with the magnetic coil idea.

We decided to cut it based on the justification that this method lagged behind the datum in the areas that we believed would really matter: reliability and accuracy.

This meant that we continued our concept generation with three concepts: the Stepper Motor / Lead Screw concept, the DC Motor / Rack and Pinion concept, and the Linear Actuator concept. Prior to continuing with concept selection, we created a rudimentary CAD model for each actuation method.

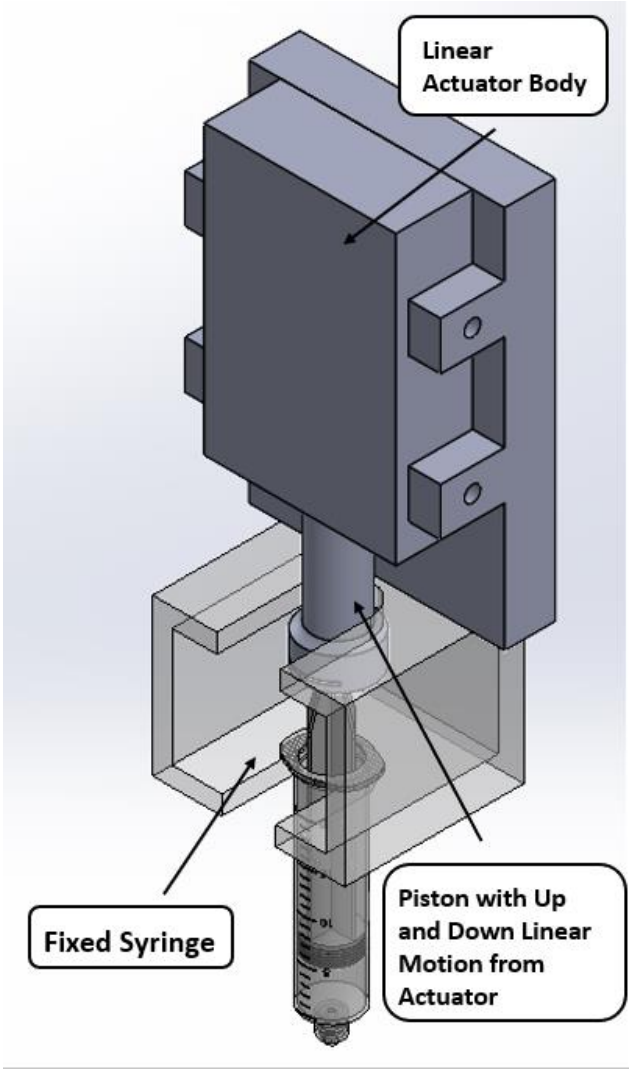


Figure C1: Rudimentary CAD for Linear Actuator

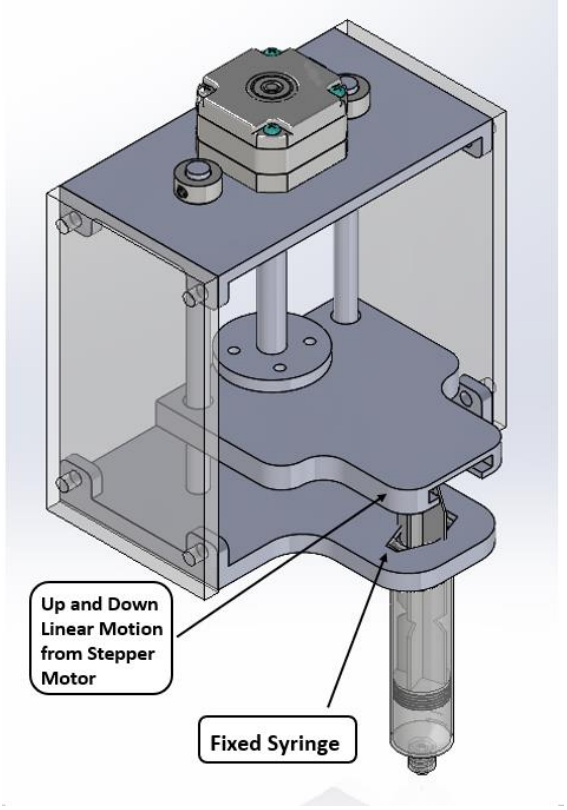


Figure C2: Rudimentary CAD for lead screw mechanism

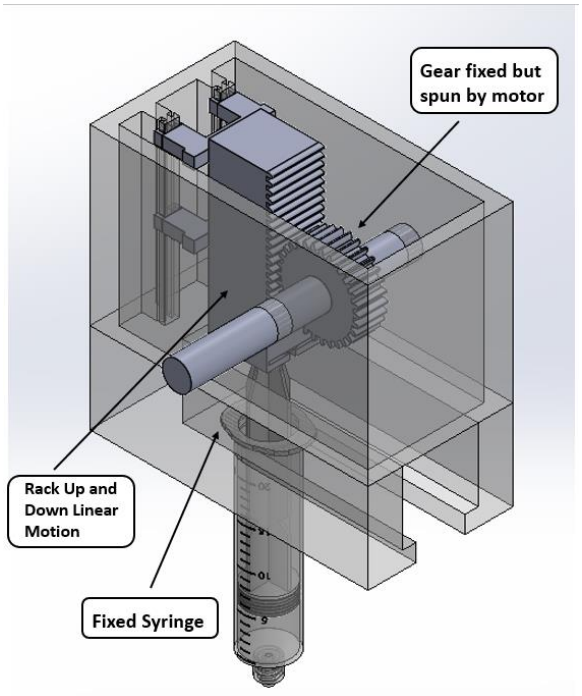
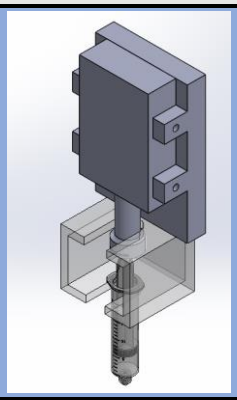
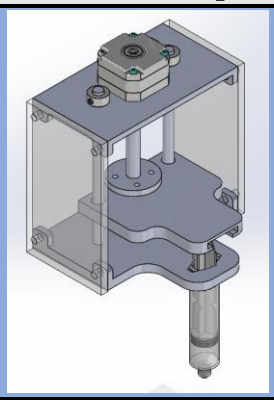
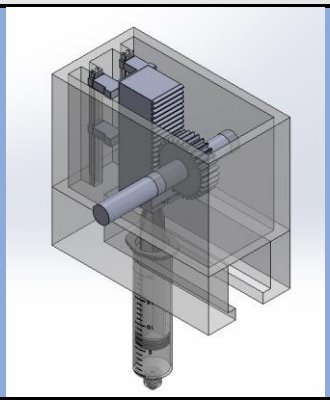


Figure C3: Rudimentary CAD for rack and pinion mechanism

To establish weighting values for each criterion, each teammate individually weighed each criterion. We averaged the weighting values of each criterion, rounding each value to the nearest 5%.

To generate the weighted decision matrix, we performed a deep dive into how these concepts will meet each criterion and created a CAD model for each concept. Based on this deep dive, we provided a more accurate rating value for each of these categories.

Table C2: Weighted Decision Matrix for Actuation Concepts

		Concepts					
							
		Linear Actuator		**Stepper Motor / Lead Screw**		**Rack and Pinion**	
Criteria	**Wt**	**Rating**	**Score**	**Rating**	**Score**	**Rating**	**Score**
Low Cost	15%	1	0.15	3	0.45	3	0.45
Low Bulk	10%	3	0.3	1	0.1	5	0.5
High Simplicity	10%	5	0.5	2	0.2	2	0.2
High Durability	15%	4	0.6	3	0.45	1	0.15
High Reliability	20%	3	0.6	5	1	1	0.2
High Accuracy	30%	3	0.9	5	1.5	1	0.3
Total		3.05		3.7		1.8	
Rank		2nd		1st		3rd	

Cost Analysis:

The cost analysis depends on the three actuation strategies, motors, and other parts that are necessary for device. Generally, the more accurate the device is, the higher the cost. We will provide the market price for each actuation technology as a reference. To do this we drafted a theoretical BOM for each concept. This BOM only accounts for actuation specific purchases.

Linear Actuator – Linear actuators have the highest cost among the three strategies. The cost ranges widely from 40 to 200 dollars each with different accuracy. We selected a mid-range actuator that would be capable of providing the necessary force to push the syringe.

Table C3: Theoretical BOM for Linear Actuator Concept

Item	Cost ($)
Linear Actuator	68.69
Total	68.69

Stepper Motor / Lead Screw – Most stepper motors move 1.8 degrees per step with a standard accuracy of ±5 percent. The commercially available stepper motor's prize ranges from 9 to 14 dollars each. Compared to linear actuators, stepper motors are much cheaper. With other parts, the total cost for this actuation method is around $50.

Table C4: Theoretical BOM for Stepper Motor Concept

Item	Cost ($)
Shaft Coupler	7.79
Linear Motion Kit	18.99
Lead Screw and Nut	9.99
Nema 17 Stepper Motor	12.98
Total	49.75

DC Motor / Rack and Pinion – DC motors' prices depend on the voltage and rpm. Generally, a higher voltage leads to both higher rpm and cost. Compared to the linear actuator, the cost of DC motor is significantly lower. Our selection for the motor is the following 12 V DC motor:

Table C5: Theoretical BOM for Rack and Pinion Concept

Item	Cost ($)
Linear Motion Kit	18.99
Rack and Pinion Kit	16.99
DC Motor	8.99
Total	44.97

Size Analysis:

To do the size analysis, we simply created a rough CAD model of each of the top three solutions. We then utilized the measure tool to obtain dimensions for each concept.

Linear Actuator – The linear actuator design has dimensions of 6″ by 4″ by 2″. This component is a placeholder spot for the parts that are required to make the actuation system function.

Steeper Motor / Lead Screw – The steeper motor/ lead screw concept consists of the largest body compared to the other two designs it has dimensions of 6″ by 6″ by 6″.

DC Motor / Rack and Pinion – This design consists of two compartments. The top compartment has the dimensions 3″ by 2.5″ by 4.8″ which stores the whole rack and pinion system.

Simplicity Analysis:

To perform the simplicity analysis, we first looked at what manufacturing strategies would be necessary, we then looked at how many parts each design utilized. These two numbers provided us with our baseline for simplicity.

Linear Actuator – The linear actuator concept will be reliant upon exclusively 3D printing and hand assembly. Since the linear actuator can be bought as a single item, it takes out a lot of the complexity of obtaining linear actuation. For this reason, we think a valid strategy could be generated using less than 8 components per pump. These make this design rank very highly in terms of simplicity.

Stepper Motor / Lead Screw – Similar to the linear actuator, the stepper motor / lead screw concept can be made using exclusively a 3D printer and with hand assembly. The hand assembly will likely be more complex because the stepper motor / lead screw concept will require a lot more pieces and will most likely require less than 15 components per pump. It will have a lot more parts that could cause a critical failure. Unlike the linear actuator, the lead screw, nut, and guide rails are all exposed and are therefore more likely to suddenly fail. For these reasons, this concept does not rank very highly in terms of simplicity.

DC Motor / Rack and Pinion – The rack and pinion method will also be able to be largely 3D printed with hand assembly. Similar to the lead screw method, this concept will require around 15 components per pump. It will also have a similar assortment of parts that could lead to critical failure: such as the guide rails, the pinion gear, and the gear rack. While each of these parts could be 3D printed, it is likely that some components, such as the gear rack, will need to be purchased to ensure high quality parts are used.

Durability Analysis:

To perform the durability analysis, we considered two factors. The first was the replaceability of the parts, and the second was how durable we predicted each mechanism to be (i.e., are there critical parts that are 3D printed, what loads will the mechanism be required to withstand).

Linear Actuator – Most linear actuators are driven by stepper motors. The actuator we have identified as competitive for this problem, the Actuonix L16 Series, produces a push force of about 200 N. This means that the entire assembly must be able to be continually loaded and unloaded with a 200 N force. This force should be easily handled by a frame, assuming it has some metal support element (i.e., bolts that run through it).

The linear actuator is also quite effective because the entire mechanism is purchased from a third party. This means that many critical components are machined instead of 3D printed, helping the overall durability.

However, if any piece in the actuator breaks, the entire thing will have to be swapped out. This will be quite expensive and the time frame for this will be much longer compared to other concepts. For this reason, despite the fact that the this is undeniably the sturdiest of the solutions, we rated this strategy a 4 out of 5.

Stepper Motor / Lead Screws – The torque produced by the NEMA 17 stepper motor is approximately 0.45 Nm. If this is run with a lead screw with a lead of 2

mm, this torque will convert to a push force of about 225 N. This will produce negligible stresses on a steel lead screw.

The primary concern with this approach is that each component will need to either be purchased or printed. Anything that is printed will be easily replaceable but is far more likely to break; the opposite is also true. A lot of critical components will also be more exposed with this method compared to the linear actuator method. This means that they are likely more prone to failure.

For these reasons, we rated the stepper motor method medium on the durability scale.

DC Motor / Rack and Pinion – The torque produced by a 12 V electric gear motor (MakerMotor: PN01007-38) is approximately 6 N-m. With a 5 mm pinion gear (a smaller size is better for accuracy), this will convert to a push force of 2400 N. While this may seem excessive, to reach an acceptable accuracy, the gear ratio must be incredibly high. This is much larger than the other actuation methods and is a concern for anything that is 3D printed. To repetitively actuate the plunger, multiple metal supports will likely need to be included in the design. The push force could be reduced by utilizing a larger pinion gear, this would cause the accuracy to decrease.

In addition, it is a concern that the components will need to be individually purchased or printed. Similar to the stepper motor, this is good because it means technicians will only need to replace parts as they fail, but it is bad because parts will generally be of poorer quality, and potentially be more expensive.

Since this actuation strategy performed poorly in both areas of performance, we rated this concept poorly.

Reliability Analysis:

There is a measurement related to reliability: mean time between failures, or MTBF. It is the average amount of time between failures for a machine or component, and that includes electric motors. An electric motor that is reliable will have a long MTBF (a long time between failures). The lifespan of the device heavily depends on how to use the device. For example, the same device will have a shorter lifespan if it is used continuously under high RPM. Therefore, we provided the mean time between failures, MTBF, as a reference for our reliability analysis.

Strategies	MTBF (hr.)
Linear Actuator	5000/20000 [2]

Stepper Motor	20000 [3]
DC Motor	10000/50000 [4]

These values for MTBF are dependent on a variety of factors, but we found sources to that cited an approximate MTBF for each of these methods.

Accuracy Analysis:

From our calculations shown in Appendix B, we have proven that the accuracy in the volumetric output will be directly proportional to the positional accuracy of our actuation strategy (this is a simplistic model to make comparisons easier). For this reason, when selecting our actuation method, we exclusively considered the error associated with the actuation method and assumed everything up and down stream would be identical in terms of error (e.g., same syringe diameter, same plunger seal deflection, same fluid).

Linear Actuator – The error associated with linear actuators widely varies greatly with the cost of the actuator. Theoretically, we could get a linear actuator that is accurate within a couple of micrometers, however these get very expensive. To keep this discussion realistic, we limited the actuators we looked at to less than $200 (approximately double the estimated budget – this should account for any major shift in the proposed budget). Not a lot of actuators exist in this price range, but we have compiled a list shown in Table C61 below.

Table C61: Cost and Accuracy of On-Market Linear Actuators

Device (Company)	Accuracy (mm)	Cost ($)
L16-100-35-12-S (Actuonix)	0.4	90
P16-50-256-12-S (Actuonix)	0.3	100
T16-100-64-12-P (Actuonix)	0.4	115
12LF-27PT-90 (mightyZap)	0.1	150

Table C61 illustrates that these linear actuators are going to be very accurate, and would be capable of providing the required accuracy, but they will come at a very steep cost. While we were able to identify cheaper linear actuators, they generally were clearly low quality and did not come from reputable vendors or did not have an accessible data sheet. For this reason, we rated this strategy highly, but not too well because of the issues that the cost presents.

Stepper Motor / Lead Screw:

The positional accuracy of this concept can be modelled as,

$$a_x = a_\theta L$$

where a_x is the uncertainty in position, a_θ is the uncertainty in the angular position, and L is the lead of the screw we utilize. The lead screw that we are planning to utilize has a lead of $2\frac{\text{mm}}{\text{rev}}$. Looking at reasonably priced stepper motors, the NEMA 17 stands out, it is typically utilized in 3D printers and other precision devices. It is priced fairly and is 5% accurate with 1.8° steps, by plugging these numbers back into the equation above, we find that this strategy has a positional accuracy of 0.01 mm. For these simpler calculations, we are not going to worry about the accuracy of each step because the error will largely be caused by the course step sizes. This is much more accurate than the other concepts, so we have rated this very highly.

DC Motor / Rack and Pinion:

The positional accuracy for this concept can be determined using a similar formula to the equation above. Since the rotational inaccuracy should be proportional to the positional inaccuracy (assuming that the gears mesh and the tolerance of the gears is much finer than the uncertainty in motor position). Assuming we utilize a 128-pulse encoder (generally cost below $20), we will have a relative angular uncertainty of $\pm 0.49\ rad$.

$$a_{enc_{rel}} = \frac{2\pi}{pulse\ count}$$

To reduce this, we could purchase a more expensive encoder, potentially up to 256-pulses, but this will increase the cost by about a factor of 2 while halving the uncertainty as well. The conversion between rotational and translation motion can be obtained by leveraging the no slip condition,

$$x = \theta r$$

where x is the linear displacement, the θ is the angular displacement, and r is the radius of the pinion gear. This equation will also be true for the uncertainty in the angular and translational directions, assuming the uncertainty in the radius is negligible compared to the encoder.

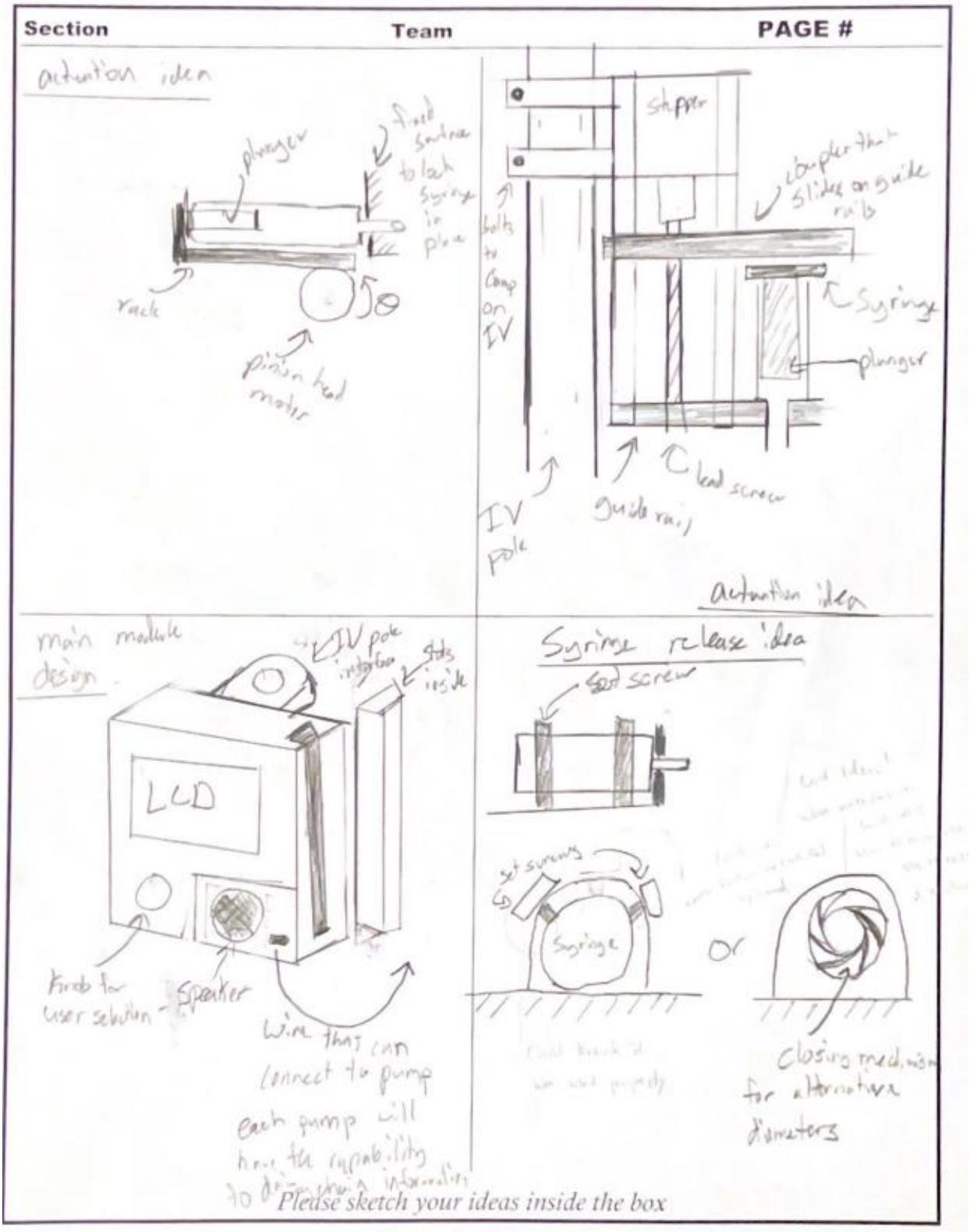
$$a_x = a_\theta r$$

Utilizing the above equation, we find that the translational uncertainty for a pinion gear with a radius of 2.5 mm (this should be about as small as we can reasonably obtain) will be about 1.2 mm. This is on par with the linear actuator, but much coarser than the stepper motor.

Based on these findings we selected the stepper motor - lead screw actuation strategy. While we considered the linear actuator method, we determined that if anything, the linear actuator's cost would be incredibly restrictive for this project. With a budget of only about $100 per pump, the

Appendix D – Full Concept Generation Results

The following concepts emerged after our first round of concept generation.



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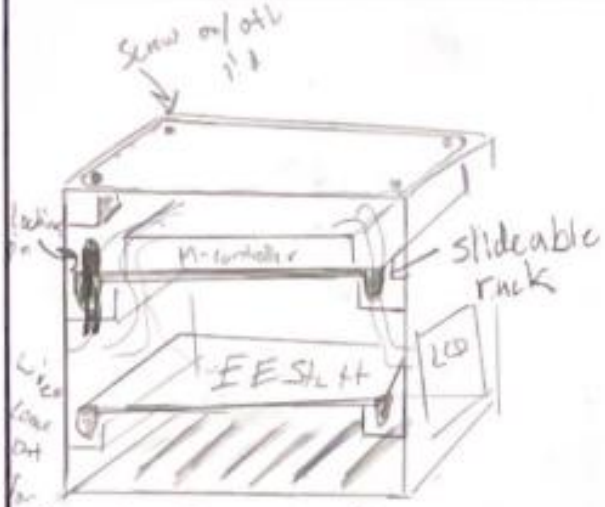

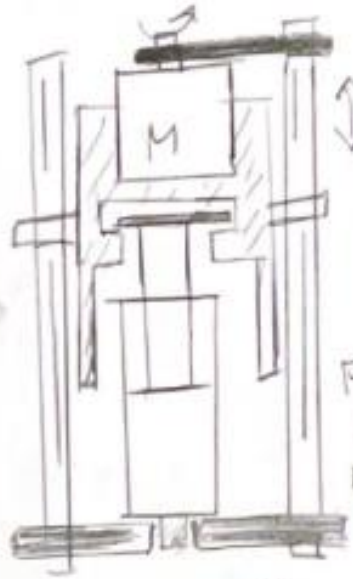

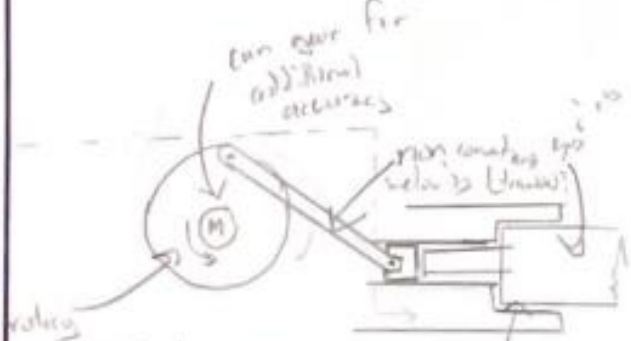

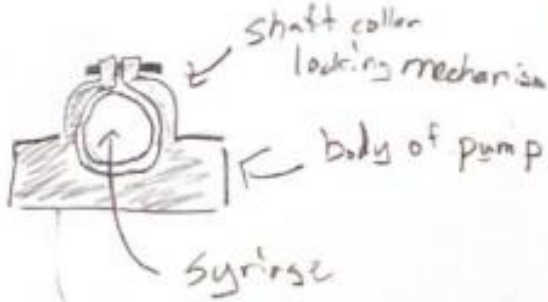

Please sketch your ideas inside the box

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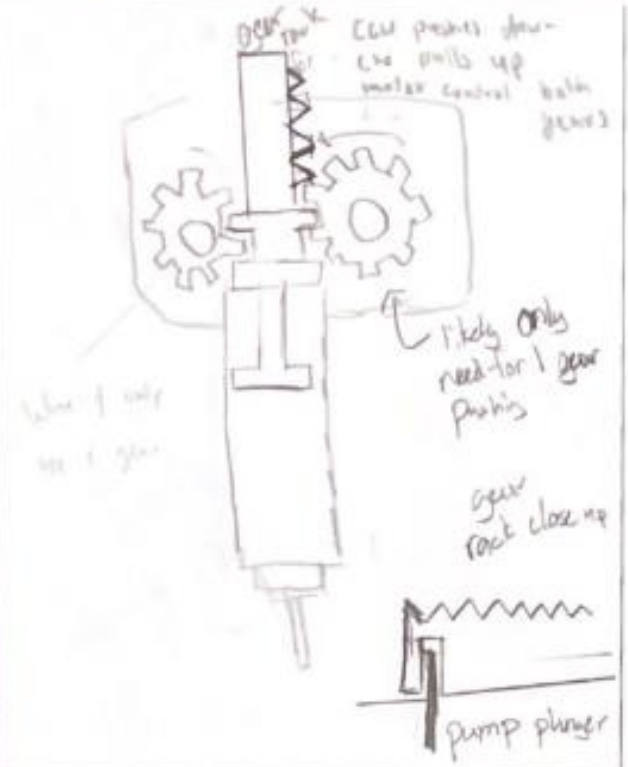

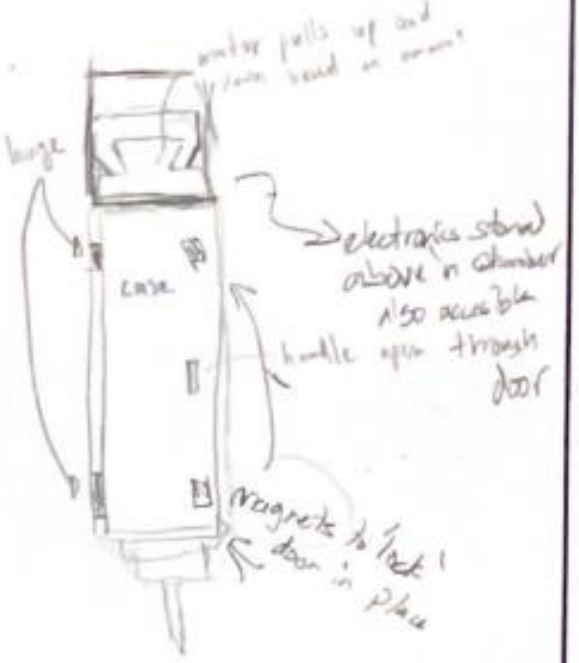

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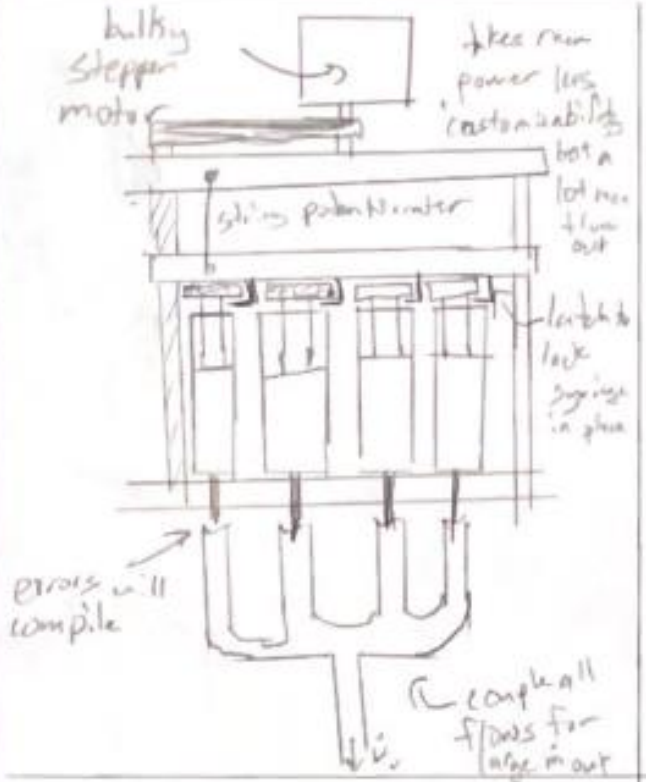

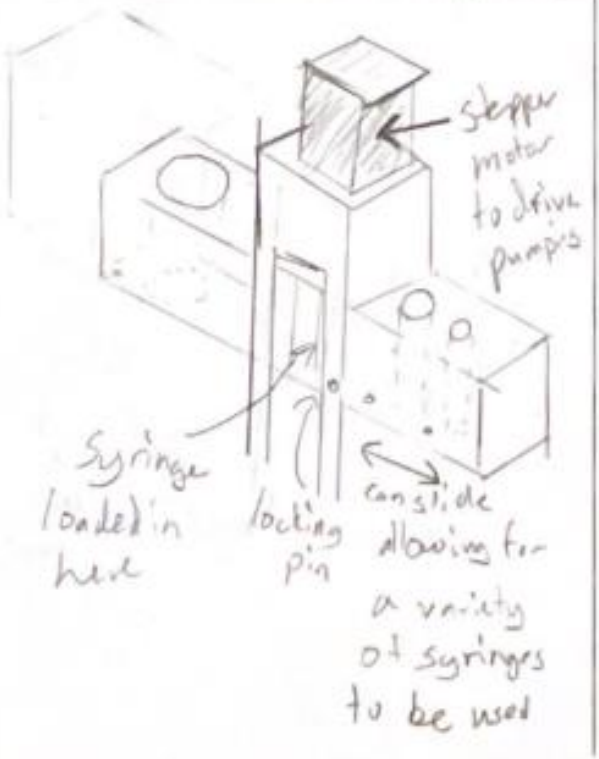

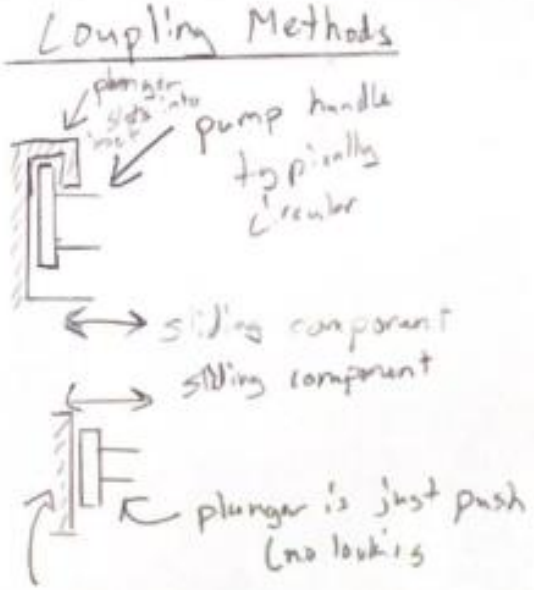

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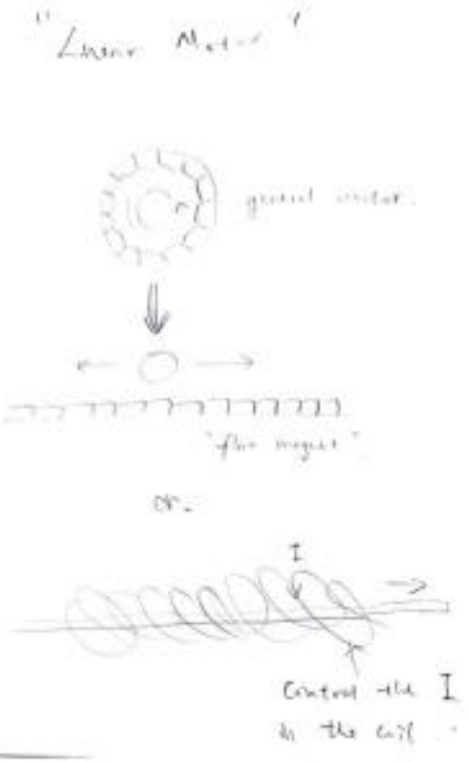
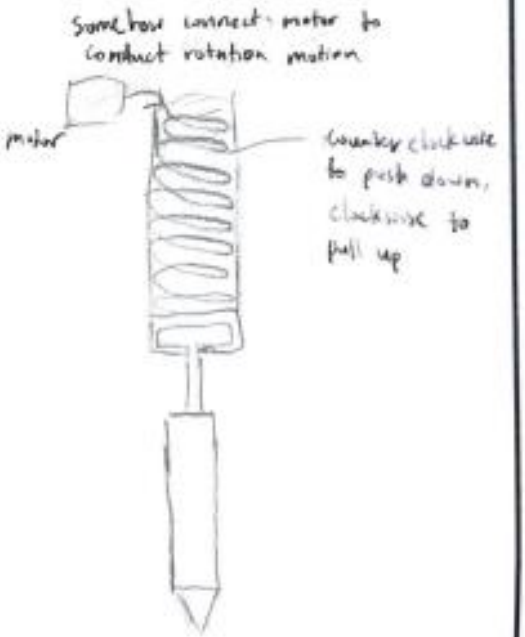
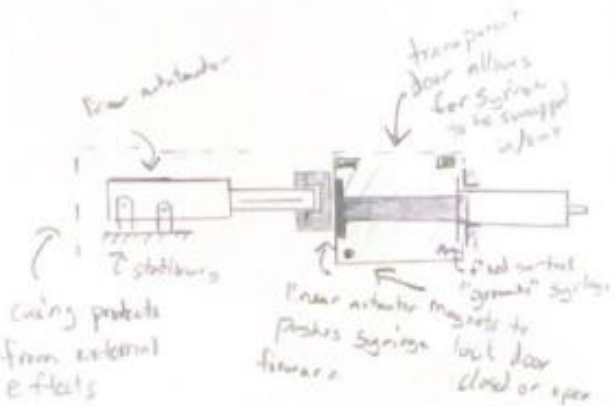


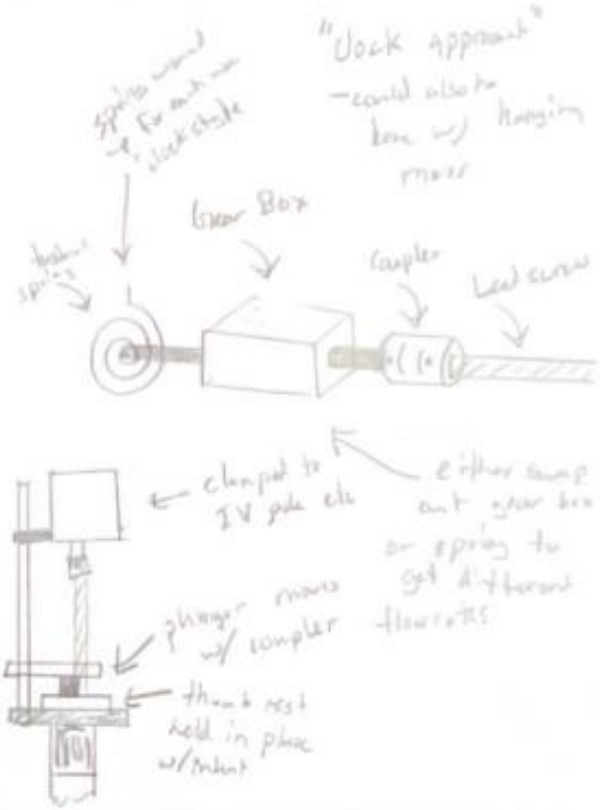
Section **Team** **PAGE #**

magnet driven linear motion



Actuation Strategy





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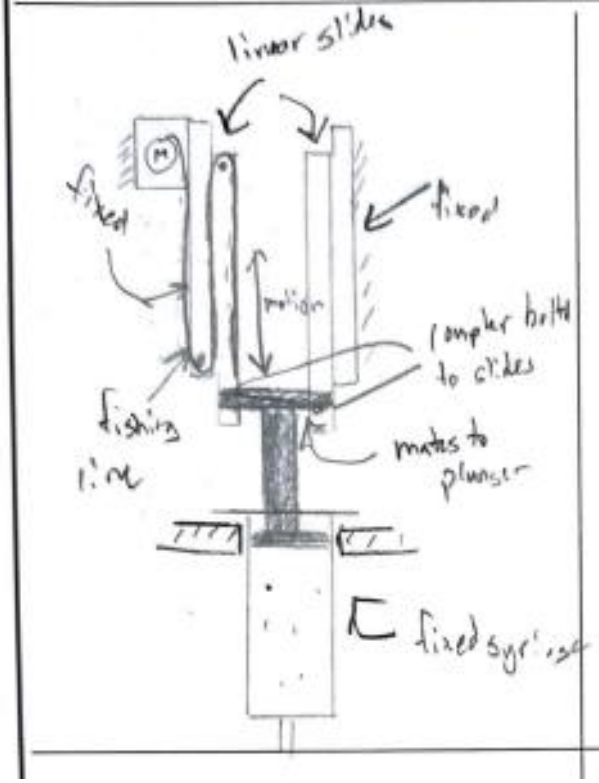


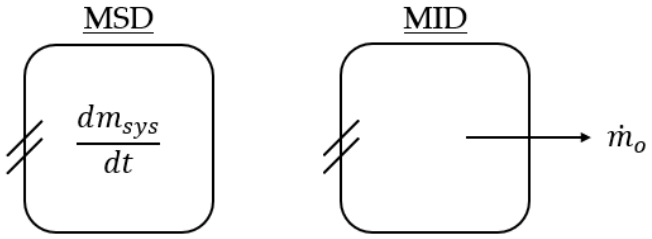
Appendix E: Conservation of Mass Model

Obtaining the model:

To start our conservation of mass model we are going to take the fluid in our syringe to be our system. We are then going to apply conservation of mass,

$$\frac{dm_{sys}}{dt} = \sum \dot{m}_{in} - \sum \dot{m}_{out}$$

Where $\frac{dm_{sys}}{dt}$ is the rate of change of mass in our system, $\sum \dot{m}_{in}$ is the sum of mass flows into our system, and $\sum \dot{m}_{out}$ is the sum of mass flows out of our system.



Based on the mass interaction diagram and mass state diagrams shown above, we can simplify this equation down to,

$$\frac{dm_{sys}}{dt} = -\dot{m}_{out}$$

where, $\dot{m}_{out}$ is the mass flow out of the syringe. If we assume that the fluid is incompressible, this can further be reduced to,

$$\frac{dV_{sys}}{dt} = -\dot{V}_{out}$$

where $\frac{dV_{sys}}{dt}$ is the volume of the system, and $\dot{V}_{out}$ is the volumetric flow out of the syringe. Assuming the syringe's body geometry can be modelled as a cylinder, allows us to alter the equation such that,

$$\frac{\pi d^2}{4}\frac{dL}{dt} = -\dot{V}_{out}$$

Where d is the syringe inner diameter, and $\frac{dL}{dt}$ is the rate of change of the length of the cylinder. $\frac{dL}{dt}$ should be proportional to the output of the stepper motor, which allows for us to further reduce the equation to,

$$\frac{-\pi d^2 l\omega}{4} = -\dot{V}_{out}$$

where ω is the angular velocity of the motor shaft, and l is the lead screw's lead. With this model, we should be able to relate a target $\dot{V}_{out}$ to an angular velocity. If we are interested in looking at the finest resolution that our syringe can output, we can integrate this equation with respect to time to find that,

$$\frac{-\pi d^2 l\Delta\theta}{4} = -\Delta V_{out}$$

with this relationship, we can see that for every step with a 50 mL syringe, we will output about 0.04 mL of fluid. (using numbers for BD 50 mL syringe and nema 17 stepper motor).

Uncertainty in Conservation of mass model:

As we are still in the early stages of this project, we assumed that the only uncertainty that would be dominant would be the uncertainty of the angular velocity. Utilizing the uncertainty propagation formula, this means that,

$$\omega_{\dot{V}} = \frac{\pi d^2 l}{4}\omega_{\omega}$$

Where $\omega_{\dot{V}}$ is the volumetric flow rate uncertainty and ω_{ω} is the angular velocity uncertainty. We hope to produce a design space to estimate acceptable ranges of lead and syringe diameters to produce the desired accuracy. As well as to see the requirement for the stepper motor's accuracy.

Appendix F – Old Design:

Ideally, we utilize pogo pins that are mounted in these slots for all data transfer between the syringe pumps and the electronics.

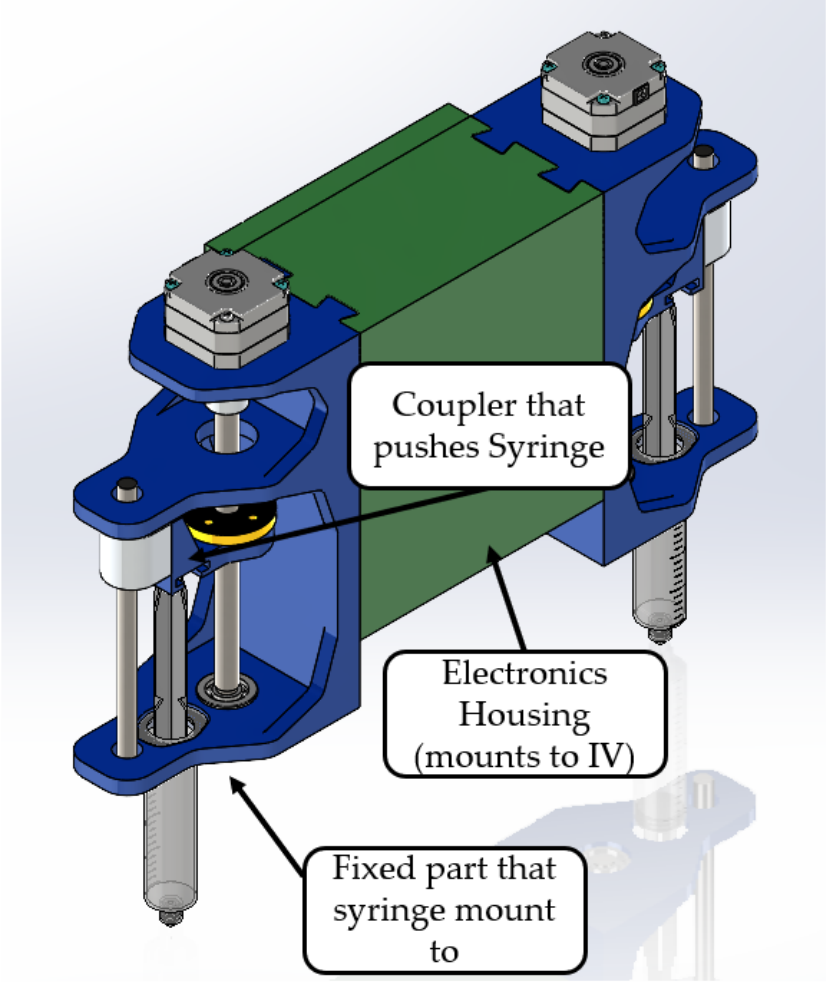


Figure F1: Design to date of syringe pumps and electronics housing.

Since the ECE team has not specified the types of user interfaces they plan on having (e.g., keypad, dial, button), we have not designed any of this. However, most likely these interfaces will attach to the front of the electronics housing. We have specifically designed it to be large so that we have flexibility for when these interfaces are given.

The syringe pump is made from two 3d printed parts, a body and a coupler.

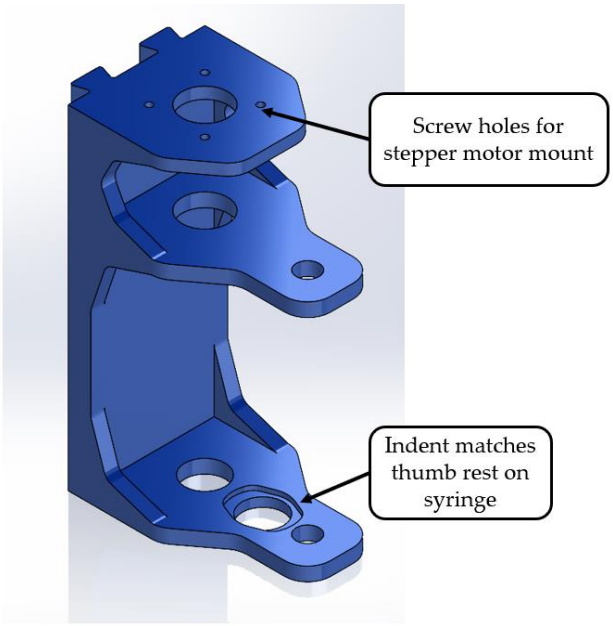


Figure F2: Syringe pump body

The body is one large 3d print that will take a long time and require support material. In the future we are hoping to reduce the bulk of this design so that it can be 3d printed faster and a print failure is less likely.

We couple the motion between the lead screw and the syringe using a handful of off the shelf components. These reduce frictional losses and will ensure smooth translational motion. However, they will be more expensive than 3d printing. Next quarter we are hoping to test this coupler with a purely 3d printed alternative.

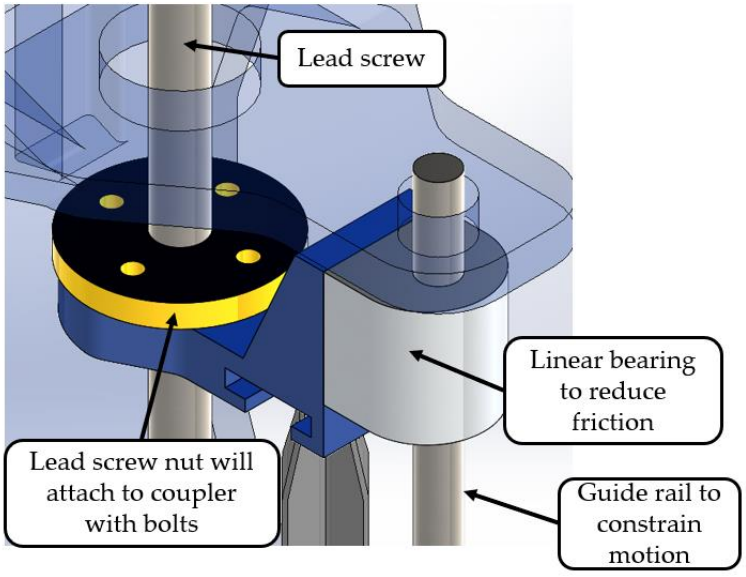


Figure F3: Coupling strategy for design to date.