Development of a Low-Cost Syringe Pump for Low-Resource Anesthesia

Akshiv Bansal
Ian Thompson

Project Sponsors:
Dr. Mark Ansermino
Dr. Guy Dumont

Applied Science 479
Engineering Physics
The University of British Columbia
January 8th, 2018

Project Number: 1752

Preface

There are no confidentiality considerations when reading this report. We thank the members of the UBC Digital Health Innovation Lab, especially Dr. Mark Ansermino, Dr. Guy Dumont, Dr. Richard Merchant, and Dr. Chris Peterson, for their support in this project.

Executive Summary Word Count: 235

Main Body Word Count: 3790

Akshiv Bansal

Ian Thompson

Executive Summary

Safe surgical care is an essential medical resource which is unavailable to billions of people in developing nations. Intravenous anesthesia plays a critical role in surgery, however conventional syringe pumps are too expensive for widespread use in low-resource settings. To address this issue, we developed a low-cost syringe pump using less than $25 of components. To meet the demands of surgery, we aimed to achieve an accuracy of ±15% at drug delivery rates of 10-200 mL/hr.

The syringe pump design employs a simple 3D printed chassis, which holds two syringes back-to-back. An air pump and valve drive one syringe, forcing anesthetic from the other syringe. Our solution substitutes expensive precision components for feedback-based control. Inlaid copper strips capacitively sense changes in the syringe volume, and a Raspberry Pi Zero uses this feedback to control pumping and deliver anesthetic smoothly. The entire system runs from a powered USB connection.

Testing over a wide range of flow rates, we found the pump met the design specifications for infusion rate accuracy. This testing employed an analytical balance to track the flow of water from the pump. These results demonstrate the viability of feedback-based syringe pump design and provide a foundation for future development. Further refinement and validation of the control algorithm will improve robustness and accuracy. Careful consideration of the surgical environment and consultation with healthcare providers will help ensure successful deployment of the device in field trials.

Table of Contents:

List of Figures

List of Tables

List of Abbreviations:

FPGA	Field Programmable Gate Array
DHIL	Digital Health Innovation Lab (Project Sponsors’ Lab)
IC	Integrated Circuit
RPi0	Raspberry Pi Zero

Introduction

Background and Significance

Anesthesia is an essential medical service, which induces paralysis or unconsciousness in the patient, allowing for easier surgery and alleviating pain for patients. Intravenous anesthetics are administered through injection into the patient. They are common and well-understood. Conventional medical practice in developed nations employs a motor-driven lead screw to force a precise flow of anesthetic from a loaded syringe.

In Essential Surgery, the International Collaboration for Essential Surgery outlines 44 procedures that would avert 1.5 million deaths every year in developing countries, many of which rely on pump-based IV anesthesia [1]. Existing commercial solutions are far too expensive for these settings, costing $1000 or more. By designing a low-cost anesthetic pump, we aim to make these critical procedures more available. Previous pump designs have showed promise for addressing this need but have been inadequate in accuracy [2], versatility [3], or total cost [4]. We believe that an innovative, closed-loop approach will allow us to address the need for accuracy while maintaining a low device cost.

There are unique challenges we faced when designing a syringe pump for use in developing nations, and constraints of the low-resource surgical environment have informed our project goals and design decisions. Our primary constraint was cost. We focused on using commonly available components and simple methods of fabrication and construction. We ensured that the system will be cost-effective at initial deployment by choosing the most effective design for a scale of tens to hundreds of pump units.

Another constraint is the lack of reliable, clean power in many surgical environments. We designed our system to be powered from a mobile phone, which is a widespread and reliable source of power. We designed our pump to have low latent power consumption, and to pull no more instantaneous power than a typical mobile phone can provide. We considered other variabilities in operating environments. Finally, we aimed to create an open source and easily adoptable design. We created simple designs using common tools and programming languages, with the intent of making build plans available to groups who want to use the device. Versatility in syringe size was considered to allow adaptation to different medical procedures.

Project Objectives

1. Design of a syringe pump for approximately $25 in components or less
2. Demonstration of a syringe pump, which delivers anesthetic at a specified flow rate between 10-200 ml/hour with an accuracy of ±15%.
3. Design of an electrical power and user interface system to allow the pump to be operated and powered directly from an Android mobile phone.
4. Design of mechanical and electrical systems to ensure the syringe pump is operable without expertise, while being tolerant to light impacts, dust, water, and variations in the environment.
5. Appealing implementation of the system so it can be demonstrated to potential sources of funding or other parties interested in supporting the project.

Report Scope and Organization

To effectively deliver anesthesia, we sought to build a closed-loop, pneumatically-actuated pump, with a capacitive feedback mechanism. We started the project with a syringe pump concept, which worked using an FPGA and a specific syringe model. Throughout the project we: designed and built a new chassis to work with any generic syringe, created a new electronic measurement and control system using feedback electronics and a Raspberry Pi Zero computer (RPi0), and validated the pump performance at a variety of flow rates.

The project excluded consideration of alternative feedback mechanisms, controllers, and actuation methods. These were excluded because of time constraints and increases to the final cost of the device. For example, the FPGA was abandoned because at the scale of our initial device production, the chip setup would be more expensive than the RPi0 solution. These topics will not be addressed in this report.

Discussion

Methods

System Design

The system is composed of three subsystems: a RPi0 controller, an air pump/valve pneumatic actuation system, and a capacitor-controlled oscillator for feedback. The feedback system is used to obtain information on the syringe position, which is processed by the RPi0 controller to generate pneumatic correction to achieve proper drug delivery rates. The pneumatic corrections are applied by actuating the air pump/valve system.

These systems are all housed in a single 3D-printed chassis to create a clean and compact prototype. Power to run the device is provided by an Android phone. All aspects of the system are intended to limit upfront cost and allow for use with minimal additional materials. The intention is to deliver a kit with everything required to run the pump out of the box, save for a soldering iron, medical tubing, a syringe, and a smartphone. Pneumatic actuation and capacitive feedback help to reduce the material cost significantly by alleviating the need for precision actuation components. Cost was the driving factor many design decisions, including selection of all electronic and electromechanical components. The total cost of the project is estimated to approximately \$23.50. The RPi0 enables ease of development in C and allows a wider open-source community to adopt and change the project.

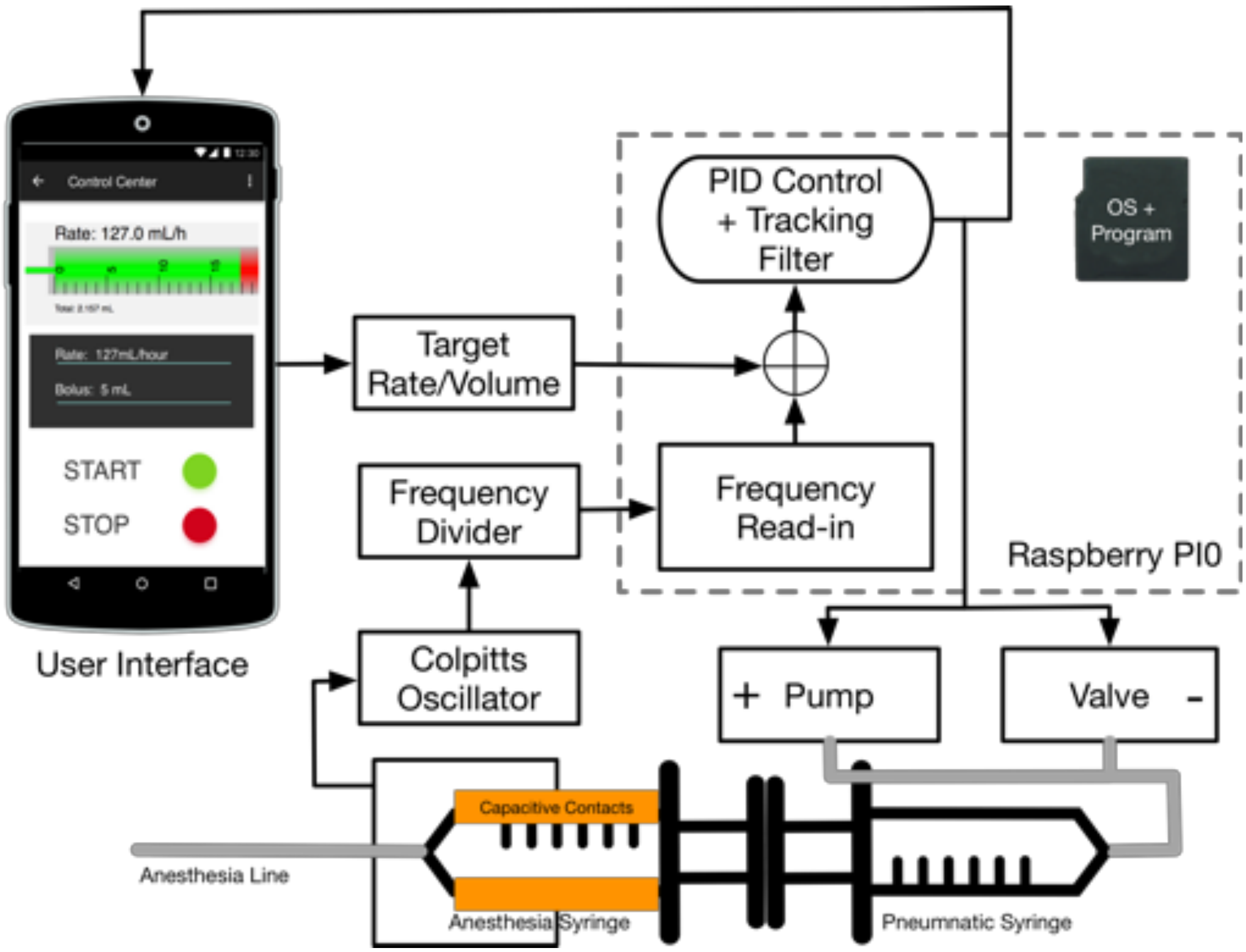


Figure 1: System Overview for Feedback and Control

Using the smartphone, the user inputs a target infusion rate and an overall volume. The rest of the system responds to deliver anesthetic as specified. The Raspberry Pi0 takes user input and issues signals to the pump/valve combination, driving the passive pneumatic syringe plunger forwards. The pneumatic syringe presses the plunger of the active anesthesia syringe, which dispenses the drug solution through the anesthesia line. Copper contacts on either side of the syringe form a capacitor, which set the frequency of a Colpitts oscillator. This signal is frequency divided and read into the Raspberry PI0 as feedback for a PID control algorithm, which in turn actuates the pump/valve system

Mechanical Design

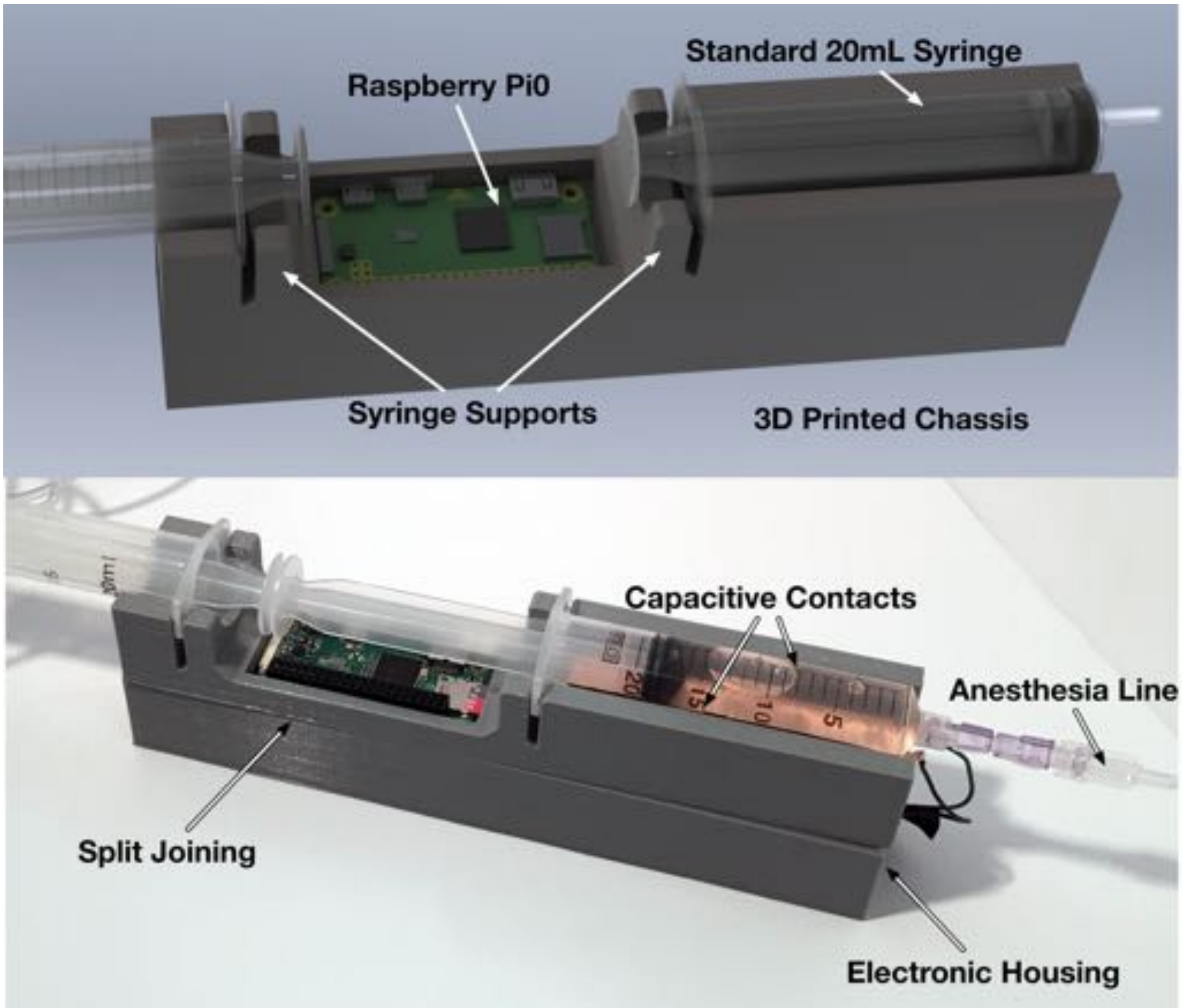


Figure 2: Mechanical Chassis Design

The chassis design is based on original work done by the project sponsors. This chassis has built in capacitive contact pads, such that generic anesthetic syringes can be used, and a clean surgical setting maintained. All of the electronics are housed inside the chassis, and the Raspberry Pi0 sits below the syringes (normally covered). The split in the prototype was introduced to simplify 3D printing, and the two halves are joined. The syringe supports are necessary to deal alternating stress from syringe installation and operation.

The chassis is designed to house all of the components and provide resistance to impact. To this end, the design focused on making an economically printable enclosure, with a setup to allow all electronics to be installed in the body, without need for electronics access or adjustments during operation.

Electrical Design

To accurately detect small capacitance changes due to syringe actuation, we constructed a capacitance-controlled variable frequency oscillator. We selected the Colpitts oscillator topology (shown in Figure 3), which produces large variations for small capacitance changes, and uses readily available, affordable components. Each oscillation must be accurately detected by the RPi0, thus we tuned component values to adjust the signal amplitude and frequency to match the RPi0’s capabilities. We found a maximum detectable frequency of 18kHz for our read-in method on the RPi0, limited by I/O bus speed. A two-stage, 18-bit frequency divider reduces the oscillation frequency to approximately 100 Hz, lowering the computational demand with minimal component cost.

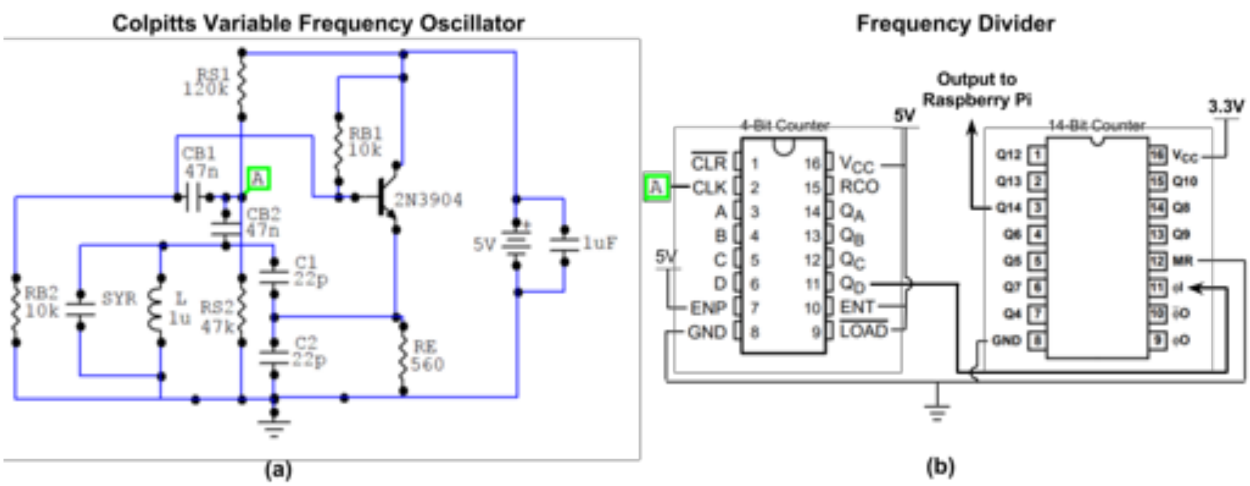


Figure 3: Electrical schematic layout of the Colpitts variable frequency oscillator used for obtaining capacitive feedback

(a) A variable frequency colpitts oscillator topology is used to create a sinusoidal oscillation at point A (labelled with a green square). Variations in the syringe capacitance (labelled SYR in the diagram), change the oscillation frequency. The frequency is in the range of 30-50 MHz, set by L, C1 and C2. We selected setpoint-biasing resistors and bypass capacitors (RS1, RS2, CB1 and CB2) to center the oscillation around the correct voltage for input into a frequency divider circuit. **(b)** A frequency divider circuit uses a 4-bit and a 14-bit counter to divide the oscillation frequency, allowing for output to the Raspberry Pi 0. Separate V_{cc} for each counter are selected to ensure the correct output voltage range.

The air pump and valve are operated using general purpose input/output (GPIO) pins on the RPi0. Since the GPIO pins have a output current lower than needed to drive the system, we constructed a simple switching circuit using an NPN Darlington transistor to power the components from the device power rails. The GPIO pins control switching; for logic high, the driver circuit powers the air pump or valve with 3.3V, and for logic low, the component is switched off.

Flow Diagram / Algorithms

The pump control is code executed on the RPi0 and is initialized by the user providing a target flow rate and total volume. Feedback from the syringe is monitored using an interrupt routine, which counts oscillations for a developer specified counting interval. Shorter counting intervals can be selected to increase response time, while longer intervals improve noise rejection. As readings are constantly generated by the feedback system, these values are compared to user targets for flow rate.

The control system actuates the valve and air pump in response to error, increasing or decreasing the air pump driving voltage if the syringe is lagging or leading respectively, or opening the valve when the syringe is moving too fast. The developer can specify weights for proportional, integral, and derivative error, as well as the threshold for releasing pressure through the valve. These parameters can be tuned to optimize system performance. The syringe pump runs until the feedback indicates that the system has infused the total amount desired, at which point the pump stops and releases all pneumatic pressure. During the pumping process, the feedback readings, pumping status, and applied valve and air pump corrections are reported for analysis. The code is implemented in C for speed and ease of execution on the RPi0.

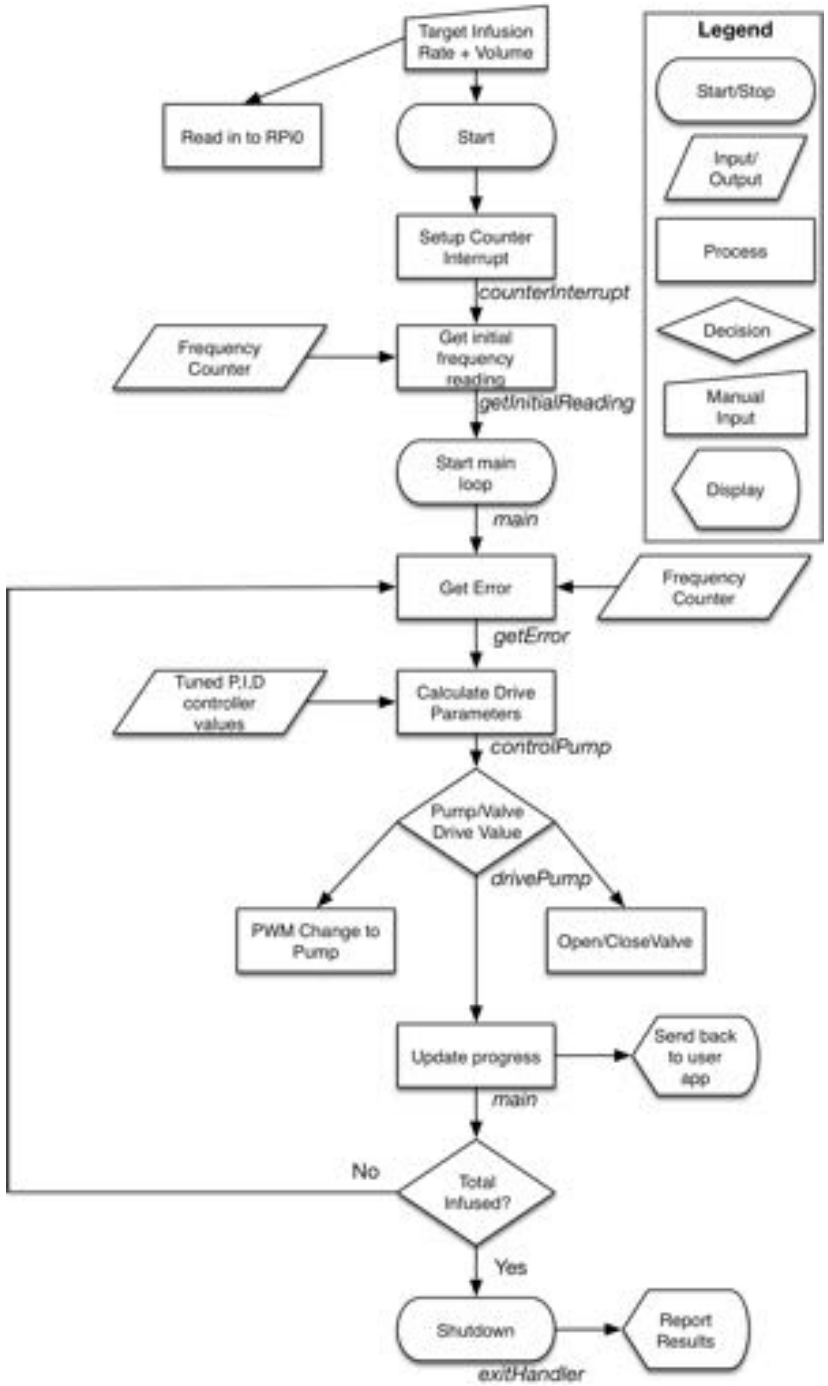


Figure 4: Software Control Flow Chart

The software implementation on the Raspberry Pi0(RPi0) handles pump control, with an Android app serving as the user interface. After the user sets the target rate and volume, the RPi0 initializes its I/O ports, and starts to read in frequency information from the oscillator/capacitor feedback system. The system stays in its primary loop until the desired amount of drug has been infused. In the loop, the system generates an error from the difference in the current syringe position and the desired syringe position. It then uses this error, along with control parameters, to drive the pump faster or slower, release the check valve, or wait until the next update. Upon reaching the volume target or encountering an error, the system goes to the exit handler, which stops all parts of the system before reporting the result to the operator and shutting down the system.

Experimental Equipment & Testing Protocol

We characterized the Colpitts oscillator output using an oscilloscope, powering the circuit from the RPi0. Capacitive feedback linearity data was obtained using the same frequency counting routine as the pump control software.

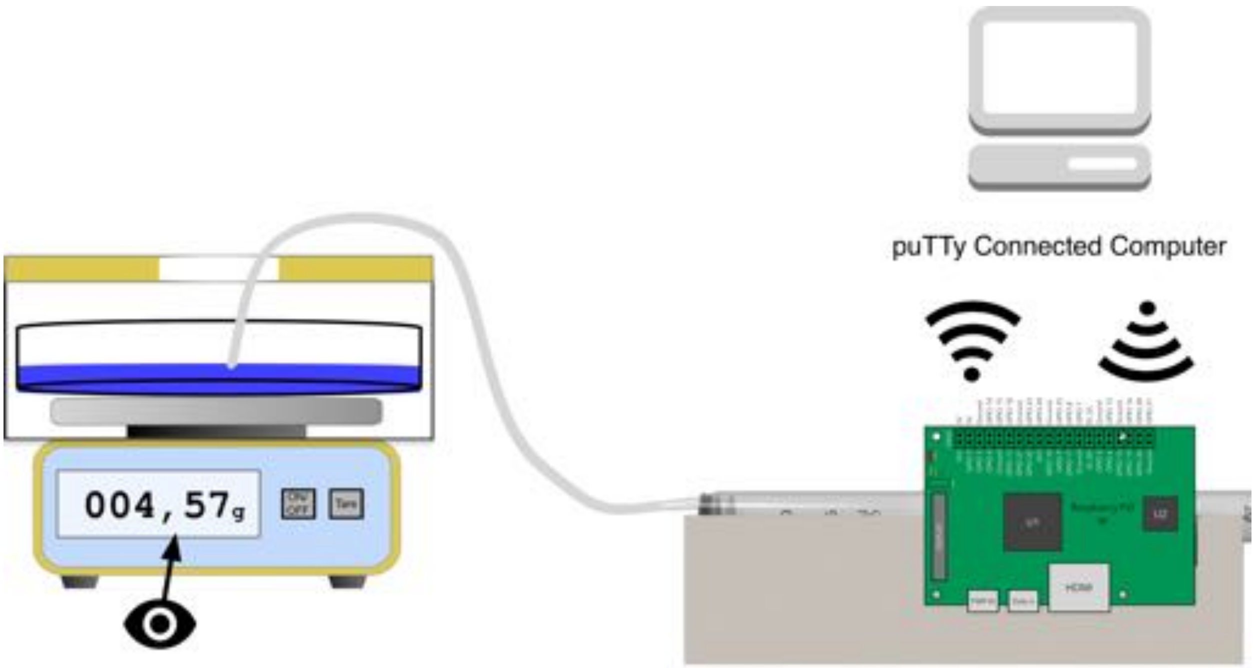


Figure 5: Testing Setup for evaluating pump performance

We used an analytical balance to test the infusion accuracy of the syringe pump. A computer linked to the Raspberry Pi Zero (RPi0) via an SSH connection was used to issue pump commands. We tested multiple infusion rates and volumes, monitoring the progression of each infusion. We set the pump up to drip through medical tubing into a plastic dish on top of an analytical balance (Sartorius BP110). To ensure minimal evaporation and variability we zeroed the balance with 20 mL of water already in the dish and used a plastic cover to minimize airflow. The data from the display on the balance was manually recorded at 1-minute intervals. All components were powered using AC wall power with a common ground.

We took pump performance data using an analytical balance to best replicate the ISO standard testing protocol (detailed in *Appendix: ISO Standard for Syringe Pumps*). This setup is less rigorous than the standard requires but provides a useful benchmark with the apparatuses we had available.

Results

The output signal from the finalized Colpitts oscillator design is shown in Figure 6 along with the frequency divided square wave, which is input to the RPi0. The circuit was powered from the RPi0, and waveforms were collected for a typical syringe loaded with 20 mL of water. We designed the Colpitts oscillator with small capacitive and inductive components to increase the sensitivity to changes in the syringe capacitance. This resulted in a high oscillation frequency of 43.91 MHz measured at the oscillator output (top subfigure). This waveform is nearly sinusoidal, with some distortion likely due to secondary resonances in the circuit. Our frequency divider was successful in generating a low-frequency output (bottom subfigure) to trigger the RPi0, reducing the oscillation to 116.4 Hz and rectifying the signal to a 0-3.3V square wave with minimal distortions or noise.

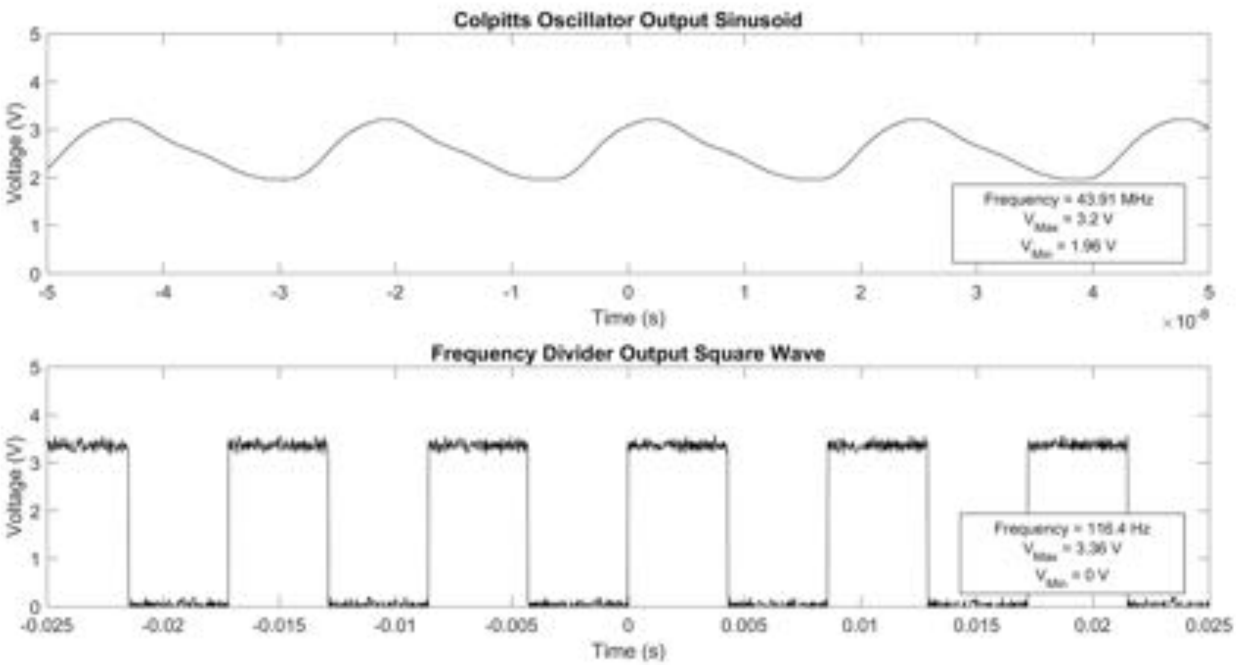


Figure 6: Oscillator Output Waveforms
We collected output waveforms of the Colpitts oscillator before and after the frequency divider using a fully loaded syringe. *Top:* The signal measured at the output of the Colpitts oscillator before the frequency divider (position A in Figure 3). The initial circuit has a high frequency (43.91 MHz) output sinusoid, with a small voltage swing. *Bottom:* The signal measured at the final output to the RPi0. The frequency divider circuit amplifies the low-voltage sinusoid to a 0-3.3V logic signal and reduces the frequency to 116.4 Hz so that it can be properly detected by the RPi0.

To simplify the controls, we assumed the changes in feedback readings would change linearly with the fluid volume in the syringe. To validate this assumption, we manually varied the volume of fluid through the functional range of 20mL syringe, sampling the feedback readings using the

RPi0. Performing a linear fit in MATLAB (shown in Figure 7) we find that the linear approximation matches our data well within our desired pumping accuracy. Over the 20 mL range of the syringe, we note a 1.84% change in the feedback readings.

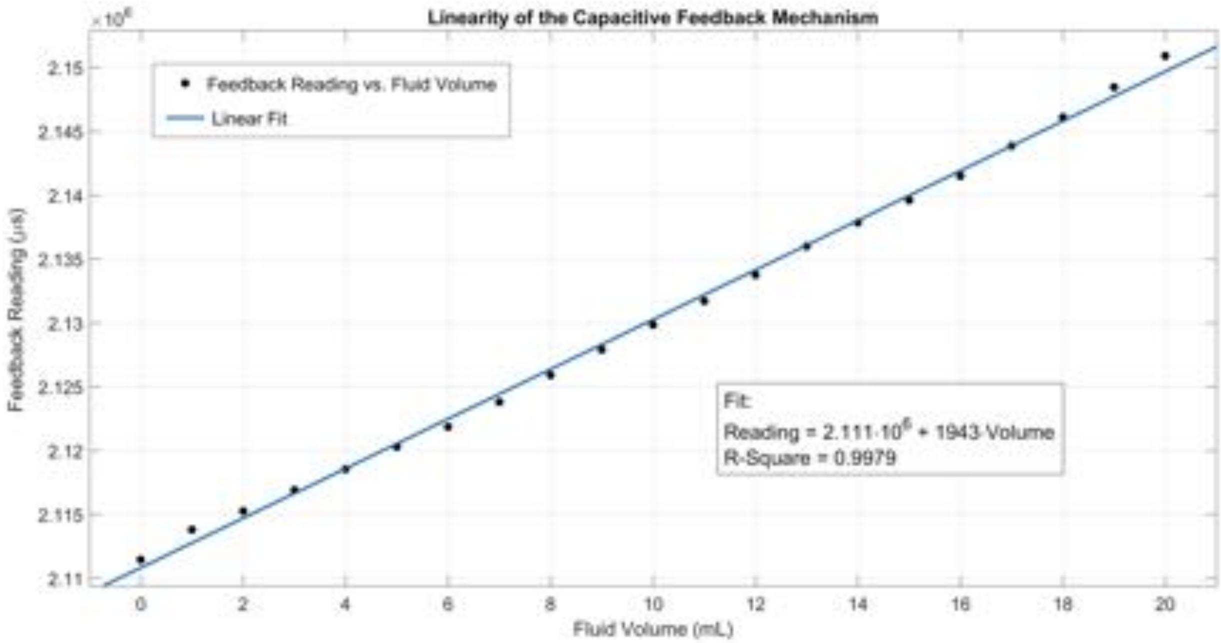


Figure 7: Capacitive feedback response linearity

We obtained feedback readings by measuring the output period of the Colpitts variable frequency oscillator across the range of syringe fluid fill volumes. A linear fit approximates the data with an R-Square value of 0.9979, demonstrating that the feedback response is sufficiently linear over the desired operation range. This linearity can be used to simplify our control scheme and accurately approximate volumes and flow rates delivered by our system.

Using our linear feedback mechanism, we tested our PID control scheme at a variety of target flow rates, following the procedure laid out in *Experimental Equipment & Testing Protocol.* Figure 5 shows the pumping results when the pump was configured to deliver a specified total volume of water, stopping when it sensed it had pumped that amount. The control system ran with only proportional control, a limit of ~40% on the maximum pumping power, and a feedback update period of approximately two seconds. At almost all time points, the pump delivered the correct volume to within ±15% (as shown by the dotted bounds in the figure). It monitored the total volume infused and stopped within 15% of the final target in 3 of 4 trials. During all trials, the pump ran independently with no operator intervention.

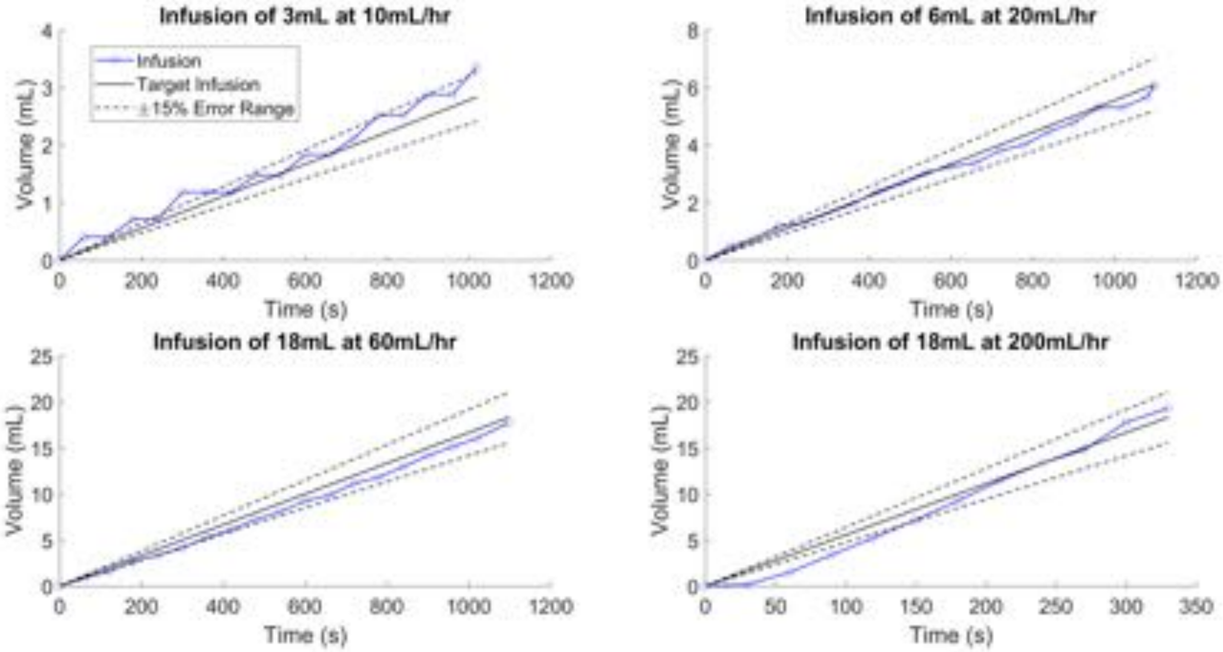


Figure 8: Syringe pump test results for a variety of flow rates

The accuracy and smoothness of delivery for our syringe pump was tested by pumping water onto an analytical balance at a range of specified rates until a specified target volume was reached. The pump was run using proportional control, updating the control response every ~2 seconds. Flow rates of 10, 20, 60, and 200 mL/hr were selected to be representative of the full range of flow rates we expect our pump to deliver. Blue lines with circles denote the resulting volume infused by our pump (calculated assuming 1g/mL water density). The solid black line denotes the target infusion rate while the ±15% volume accuracy we aim to achieve is given by the dotted lines. Our pump meets our accuracy criteria for the majority of flow rates tested, however the presence of stiction leads to rougher performance in very slow pumping, seen in the step-like plot for 10 mL/hr infusion.

Discussion of Results

The Colpitts oscillator outputs a reliable, well-formed signal to communicate feedback to the RPi0, however robustness of the output can be improved. The voltage oscillation from 1.96 to 3.2 V is sufficient to be detected as input to the frequency divider circuit, however variations in syringe capacitance or supply voltage may decrease the amplitude to under the detection threshold. It was noted in some cases that oscillations could be lost if the syringe was improperly placed or connections were loose. Modifications to buffer or increase the amplitude of the high-frequency oscillator output, to improve connections, or to shield the circuit from outside static or fields from the pump and valve may increase reliability. The counting ICs allowed for simple reduction of frequency, making the feedback signal suitable for read-in to the RPi0, however this leads to a reduction in precision when counting numbers of oscillations. We believe this approach is satisfactory for our control system. Increasing the precision may enable lower flow rates to be delivered more accurately.

We can use the linearity of our feedback system response to simply and reliably monitor the position of the syringe plunger, providing information on anesthetic flow rate and total volume delivered. However, because the total change in readings over the range of the syringe is only 1.84%, and because we often deliver quantities smaller than the 20 mL full range, the sensitivity of our feedback method may pose an issue. For very slow flow rates, the desired changes in feedback readings can be very small, with noise posing a significant challenge to precision. The 15% pumping accuracy target affords some inaccuracy; however, the noise sensitivity necessitates careful design of the control algorithm to reject noise.

The pumping performance suggests that our simple PID control algorithm is sufficient for achieving volumetric accuracy to within 15% during infusion. While the pump exceeds the error bounds at a few points, further testing and refinement of the control algorithm and parameters may lead to consistent accuracy.

The pump system appears to perform best at flow rates in the middle of the specified range of 10-200 mL/hr. Because of stiction, the pump delivered anesthetic in small discrete increments, rather than at one constant flow rate, alternating between idling and pumping somewhat rapidly. This behavior can be seen in the step-like performance at the lowest flow rate (10 mL/hr) where the quantity delivered per minute is so small that periods of idling last longer than one data point. For all tests, and especially those at high flow rates, the pump showed more significant error at the beginning of infusion, likely due to errors in estimating the initial position at startup, or delays in pumping to an operational air pressure.

Several control system changes could be tested to correct for these issues. The lengthy feedback update period of two seconds may be reduced to increase response time, leading to smaller step-like increments and overall smoother behavior. However, reducing this time may lead to increased noise sensitivity at low flow rates. Another approach to increasing smoothness may be to lower the maximum power limit, however this may limit the pump's ability to overcome obstructions in the line. A more robust solution could be varying the output power limit based on the desired flow rate, as this may compensate for both the pulsing at low flow rates and the startup inaccuracies at high flow rates. With careful adjustment and testing of the control algorithm, we are confident that we can reliably achieve our accuracy target at all flow rates.

Conclusions

We have successfully demonstrated a novel feedback mechanism for monitoring syringe position during pump operation. Using capacitive sensing and a Colpitts oscillator circuit, we generate a frequency signal, which varies with the syringe position. By demonstrating consistent linearity of the feedback response, we showed that this system can monitor and adjust the drug delivery process, increasing accuracy. This development represents a fundamental difference from other syringe pumps, which rely only on precision components and open loop control.

Our pumping approach has been implemented with a compact and easy to use design. The selection of a 3D-printed chassis allowed for easy integration of capacitive contacts, secure housing of pump components, and design reproducibility. All components were constructed with easily available parts, for a total parts cost of ~$23.50 (see *Bill of Materials* in Appendix for cost breakdown). We believe these design choices will encourage interest in our proof of concept device and attract potential funding or continued development efforts.

Our device achieved pumping accuracy to within ±15% over our desired range of flow rates from 10-200 mL/hr. This performance shows that our first proof of concept merits further development and design refinement. The main barrier to adoption of our device for clinical use is its reliability and complexity of use, and this must be addressed going forward. Adjustments to the control scheme may increase accuracy for low and high flow rates. Sophisticated knowledge of the project is needed to construct and operate the device in its current state. Our recommendations aim to finalize the device in a form more suitable for widespread use.

Project Deliverables

List of Deliverables

1. Mechanical and Electrical Design files
    a. Includes all documentation, design work, and component lists that would be needed to make the pump at scale (i.e. at the desired $25 price point)
    b. The designs are delivered in a “manufacturing ready” form
    c. Documented sufficiently to allow for greater extensions/improvements

We will deliver these files to the project sponsors. The manufacturing readiness of the project is not as expansive as originally forecasted. Additional units will still need to be 3D printed instead of injection molded. Electronics will need to be assembled on prototype PCB, not outsourced to a PCB manufacturer. All electronic design files will be available in the same public GitHub repository as the software.

2. Physical Prototype Syringe Pump
    a. Delivery of a working syringe that has a volumetric accuracy of ±15% and performs in the flow rate range: 10-200mL/h
    b. The prototype has a polished design
        - Electronics are not visible
        - Mechanical chassis is sealed
        - Design can deal with modest shocks
    c. The prototype can run off an Android phone for at least 2 hours, and is operable through simple GUI
    d. Moisture and splash proofing of the pump, including protecting electronics.

The prototype meets our specified accuracy over the desired range of flow rates. The chassis is stronger and lighter than before, all components can be nicely fit together and displayed. The chassis is not sealed or splash-proof. While the prototype will be powered by an Android phone after our ongoing commitments are fulfilled, it cannot yet be operated by one. We will schedule a handoff meeting where we will give our sponsors two 3D printed bodies, one with the working pump inside. The prototype will be setup to give demonstrations and is easily configurable for further validation.

3. Final Documentation (including Engineering Recommendation Report)

a. Includes all design documentation and methods used
b. Includes all results from prototype validation
c. Includes a list of recommended improvements before field testing

This report meets this deliverable as intended. During the hand-off meeting we will determine if an operating manual is required, and what its requirements are (discussed in *Ongoing Commitments*).

4. Access to code developed for running the syringe pump

Our open-source GitHub repository contains readable code, instructions for compiling/building, and allows for reproduction and modification.

Financial Summary

Item	Quantity	Vendor	Unit Cost	Purchaser	Note
3D Printing Services	1 Pump Chassis Print	ECE Student Services	$39.20	Guy Dumont	Sponsor's ECE Speed Chart
4-Bit Counter SN74F163AN	5	Digikey	$0.95	Project Lab	Project lab Funds
14-Bit Counter CD74HC4060E	5	Digikey	$0.87	Project Lab	Project lab Funds

Table 1: Summary of project expenses

At the resolution of the project, no funds are owed between parties. All other project materials not listed below were provided by project sponsors or the engineering physics project lab but not purchased specifically for this project.

Ongoing Commitments

1. Project debrief meeting - *Akshiv & Ian*, End date: Mid-Feb
   - We will meet with the sponsors to demonstrate our final pump performance and discuss the future of the project.
2. Establishing Android phone powering of syringe pump - *Akshiv*, End date: March 8th
   - Finish development of an app to allow the pump to run from a mobile phone. The pump should be able to run without intervention for 2 hours.
3. Creation of an operation/development manual - *Akshiv & Ian*, End date: April 1st

- The requirements will be determined in the handoff meeting. This work is intended to take a maximum of one week

4. Improvements to prototyping quality - *Akshiv & Ian*, End date: Feb 15th
   - Finishing Solidworks model to include modest sealing of chassis (12 hrs. effort)
   - Clean up internal wiring and connections (6 hours effort)

Recommendations

1. Better securing of syringes and other components

The chassis presently features no good mechanism to secure the syringes firmly into the chassis. This could be done by attaching elastic bands over the syringes or some change in the mechanical design of the chassis. Implementing this will improve the reliability of the system and allow more reliable contact between the inlaid capacitive contacts and the syringe. Further, the electronics within the chassis could be better secured to prevent shocks from affecting system performance. This could be done using adhesives, slots, fasteners, or other mechanical redesigns.

2. Further control tuning and performance validation

The pump performance provided a valuable proof of concept, but the control system needs refinement to increase reliability and versatility. We recommend testing the effects of shortening the feedback response time or adjusting the maximum pump power limit based on flow rate. These adjustments may improve the pump accuracy beyond our current limits. Repeated testing of the pump for consistency between infusions is also valuable. The pump should also be tested with a variety of outlet attachments on the end of the anesthetic line to simulate infusion in a clinical setting. Significant fluid resistance or back-pressure may be introduced when a needle, adapter, or live patient is added into the fluid line. These scenarios should be carefully considered and tested.

The robustness of the device to real-world operating conditions should be evaluated by introducing factors such as large temperature changes, vibrations, movement of the pump body, and interfering electromagnetic fields (i.e. nearby motors). A successful design will be unaffected by these factors, and these issues may necessitate mechanical or electrical modifications. Possible failure modes and misuse during operation should be investigated, such as power disconnection, loss of communication to the device, or accidental operator input. These cases may lead to considerations of fail-safes in software. Testing all these possible cases will ensure that the system is fit to be used in the clinical setting. The final validation process must be done in consultation with low-resource health care providers.

3. Mobile UI

To allow full deployment of the pump in the field, an Android app to operate the pump should be created. The scope of this project entailed creating and app to power the pump from a phone, however we did not consider phone-based operation by a healthcare provider. This user interface needs to enable the operator to set flow rates and target volumes, to start and stop the pump, and to monitor anesthetic delivery.

Figure 9: Proposed UI for Android phone development

The UI allows the operator to set the target volume and infusion rate, monitor the infusion progress, and most critically start/stop the infusion.

4. Field Validation and Scaling Up

To develop this concept further we recommend first testing the system as designed on a small scale in the field. The current approach using a RPi0 has been developed for optimal cost effectiveness at small scale. After completing the recommendations, a field test with <100 units is needed to verify the validity of using this device in a surgical setting. Once lessons learned in field deployment have informed design modifications, the project can proceed to design for larger scale. The device can be made more affordable and reliable through redesign for large scale production. This involves designing a permanent PCB with an integrated FPGA, updating the mechanical design to allow for injection molding, and creating an injection molding mold.

References

[1] Essential Surgery. Washington (D.C.): World bank group, 2015. Print.

[2] Wijnen B, Hunt EJ, Anzalone GC, Pearce JM. Open-Source Syringe Pump Library. PLoS ONE 2014 9(9): e107216. doi: 10.1371/journal.pone.0107216

[3] Mundle S, Regi A, Easterling T, Biswas B, Bracken H, Shekhavat DR, Durocher J, Winikoff B. Treatment approaches for preeclampsia in low-resource settings: a randomized trial of the Springfusor pump for delivery of magnesium sulfate. Pregnancy Hypertens. 2012;2:32–38. [PubMed]

[4] Juarez, Alexa et al. "AutoSyP: A Low-Cost, Low-Power Syringe Pump for Use in Low-Resource Settings." *The American Journal of Tropical Medicine and Hygiene* 95.4 (2016): 964–969. *PMC*. Web. 7 Sept. 2017.

[5] CAN/CSA-C22.2 No. 60601-2-24:15; Medical electrical equipment — Part 2-24: Particular requirements for the basic safety and essential performance of infusion pumps and controllers. CSA Group, Mississauga ON.

Appendices

Bill of Materials

	Component	Vendor	Manufacturer	Source	Minimum Order	Cost (USD)	Quantity	Lead Time
1	Solenoid Valve	EBay	bonk*town	A.	1	1.59	0	2 months
			maaqii	B.	1	1.63	1	1 month
2	Diaphragm Pump	Aliexpress	DIT	C.	10 Pieces	2.419	0	20 Days
			SAILFLO	D.	1	2.66	1	20 Days
3	Darlington Transistor	Digikey		E.	100	0.1713	2	
4	10uF E. Cap	Digikey		F.	1	0.22		
5	1k Resistor	Digikey		G.	1	0.1		
6	Diode	Digikey		H.	100	0.0914	2	
7	Raspberry Pi 0			I.	1	5	1	
8	2gb microSD	Aliexpress		J.	10	3.5	1	20 days
9	USB Cable			L.	1	0.6	1	40 days
10	Oscillator Circ		See below		1	2.85	1	
				Total:		16.96		
				Total (CAD)		23.50		

Table 2: General Bill of Materials

A. http://www.ebay.com/itm/Mini-2-Position-3-Way-Electric-Control-Solenoid-Valve-Gas-Air-Pump-DC-5V-6V-HM/232395417786?_trkparms=aid%3D222007%26algo%3DSIM.MBE%26ao%3D2%26asc%3D41376%26meid%3D6d133a7f9853490099fb0b4b447a6d6e%26pid%3D100623%26rk%3D2%26rkt%3D6%26sd%3D122609196665&_trksid=p2047675.c100623.m-1
B. http://www.ebay.com/itm/Mini-2-Position-3-Way-Electric-Control-Solenoid-Valve-Gas-Air-Pump-DC-5V-6V-MA-/112403400789?hash=item1a2bc3c855:g:rJAAAOSwIaFZFCDa
C. https://www.aliexpress.com/item/10pcs-New-370-DC-Air-Pump-370-motor-3V-dc-450mA-Oxygen-pump-Mini-air-Pump/32728832531.html
D. https://www.aliexpress.com/item/DC-6V-Mini-Air-Pump-Motor-For-Aquarium-Tank-Oxygen-Circulate-Durable/32830199091.html?traffic_analysisId=recommend_2088_2_90158_iswistore&scm=1007.13339.90158.0&pvid=f1234509-bb2c-48b8-8bce-fdf48264d68d&tpp=1
E. https://www.digikey.ca/product-detail/en/fairchild-on-semiconductor/KSP13TA/KSP13TACT-ND/4314808
F. https://www.digikey.ca/product-detail/en/micro-commercial-co/1N4007-TP/1N4007-TPMSCT-ND/773694

G. https://www.digikey.ca/product-detail/en/panasonic-electronic-components/ECA-1EM100B/P19522CT-ND/6109420
H. https://www.digikey.ca/product-detail/en/stackpole-electronics-inc/CF14JT1K00/CF14JT1K00CT-ND/1830350
I. Various sources Canakit, Adafruit, etc.
J. https://www.aliexpress.com/item/10Pcs-Lot-64MB-128MB-256MB-512MB-1GB-2GB-4GB-8GB-TF-Card-Micro-SD-Cards-Micro/32819257976.html?ws_ab_test=searchweb0_0,searchweb201602_3_10152_10065_10151_10130_10068_10344_10345_10342_10343_10340_10341_10139_10307_10060_10155_10154_10056_10055_10054_10537_10059_10536_10534_10533_10532_100031_10099_10338_10103_10102_10052_10053_10107_10050_10142_10051_10084_10083_10080_10082_10081_10178_10110_10111_10112_10113_10114_9998_10312_10313_10314_10078_10079_10073-9998,searchweb201603_17,ppcSwitch_5&btsid=43012fd7-ad1e-40bb-a59e-a7a1af898e6a&algo_expid=b0101ad3-e1b2-47fc-b655-c3a5fa6c4e23-0&algo_pvid=b0101ad3-e1b2-47fc-b655-c3a5fa6c4e23
K. https://www.aliexpress.com/item/Microusb-To-Female-USB-Host-Cable-OTG-Adapter-for-Lenovo-Xiaomi-Lg-Tablet-Android-Reader-Cabo/32670313423.html?ws_ab_test=searchweb0_0,searchweb201602_3_10152_10065_10151_10130_10068_10344_10345_10342_10343_10340_10341_10540_10139_10307_10060_10155_10154_10056_10055_10054_10539_10538_10537_10059_10536_10534_10533_100031_10099_10103_10102_10052_10053_10107_10050_10142_10051_10325_10084_10083_10080_10082_10081_10178_10110_10111_10112_10113_10114_10312_10313_10314_10078_10079_10073,searchweb201603_2,ppcSwitch_5&btsid=62edbde3-9e9e-4d37-ad84-00e080925017&algo_expid=828e3520-3206-4af5-9767-4ca5c9219ef5-0&algo_pvid=828e3520-3206-4af5-9767-4ca5c9219ef5

Oscillator Circuit Bill of Material

	Component	Vendor	Source	Minimum Order	Cost (USD)	Quantity
1	10k Resistor	Digikey	A.	250	0.01768	2
2	120k Resistor	Digikey	B.	100	0.0232	1
3	56k Resistor	Digikey	C.	100	0.0232	1
4	560 Resistor	Digikey	D.	100	0.0232	1
5	47nF Cap	Digikey	E.	100	0.1079	2
6	22pF Cap	Digikey	F.	100	0.0985	2
7	2N3904 NPN BJT	Digikey	G.	100	0.139	1
8	1uH Inductor	Digikey	H.	100	0.349	1
9	1uF Ceramic Cap	Digikey	I.	100	0.135	1
10	2 pos Header pin	Digikey	J.	100	0.0817	1
11	PCB (1.5"x1')	Chopped from Projectlab		1	0.5	1

12	4Bit Fast Counter	Digikey	K.	100	0.6074	1
13	14 Bit Counter	Digikey	L.	100	0.5282	1
				Total:	2.85806	

Table 3: Oscillator Bill of Materials

A. https://www.digikey.ca/product-detail/en/stackpole-electronics-inc/CF14JT10K0/CF14JT10K0CT-ND/1830374
B. https://www.digikey.ca/product-detail/en/stackpole-electronics-inc/CF14JT120K/CF14JT120KCT-ND/1830401
C. https://www.digikey.ca/product-detail/en/stackpole-electronics-inc/CF14JT56K0/CF14JT56K0CT-ND/1830393
D. https://www.digikey.ca/product-detail/en/stackpole-electronics-inc/CF14JT560R/CF14JT560RCT-ND/1830344
E. https://www.digikey.ca/product-detail/en/vishay-bc-components/K473K15X7RF5TL2/BC1082CT-ND/286704
F. https://www.digikey.ca/product-detail/en/vishay-bc-components/K220J15C0GF5TL2/BC1005CT-ND/286627
G. https://www.digikey.ca/product-detail/en/fairchild-on-semiconductor/2N3904BU/2N3904FS-ND/1413
H. https://www.digikey.ca/product-detail/en/murata-power-solutions-inc/11R102C/811-2018-ND/1998205
I. https://www.digikey.ca/product-detail/en/nichicon/UMT1H010MDD1TP/493-10286-1-ND/4312545
J. https://www.digikey.ca/product-detail/en/harwin-inc/M20-9990246/952-2262-ND/3728226
K. https://www.digikey.ca/product-detail/en/texas-instruments/SN74F163AN/296-14810-5-ND/562694
L. https://www.digikey.ca/product-detail/en/texas-instruments/CD74HC4060E/296-25998-5-ND/1507209

Obfuscated in these Bill of Materials is the cost of manufacturing a chassis, which presently is the largest cost of pump. We were able to print the chassis for approximately $40, or roughly double the sum total of all other component. We expect that with injection molding the per unit cost of the chassis will become insignificant.

ISO Standard for Syringe Pumps

There is a well-established standard for syringe pumps that govern what the international community expects from a modern syringe pump. Replicating the basic performance characteristics forms the basis of the design parameters we are aiming for. Standard IEC 60601-2-24, represents the international standard and CSA-C22.2 is the Canadian Standard Agency equivalent. The standard is a rather long document and chronicles the specifics of every aspect of any general infusion pump. Most relevant to this project is the information on volumetric flow range, accuracy, and testing requirements for said pumps.

In the following equations Ep refers to the percentage error from the desired rate.

Calculate flow using the expression:

$$Q_i = \frac{60\,(W_i - W_{i-1})}{S\mathrm{d}} \text{(ml/h)} \tag{1}$$

i = 1, 2 .. T_0/S

where

W_i is the i^{th} mass sample from the analysis period T_0 (g) (corrected for evaporative loss);

T_0 is the analysis period (min);

S is the sample interval (min);

d is the density of water (0,998 g/ml at 20 °C).

Calculate E_p(max.) and E_p(min.) using the trumpet algorithm as follows:

For observation windows of duration P = 2 min, 5 min, 11 min, 19 min and 31 min, within the analysis period T_x, there are a maximum of m observation windows, such that:

$$m = \frac{(T_x - P)}{S} + 1 \tag{2}$$

where

m is the maximum number of observation windows;

P is the observation window duration;

S is the sample interval (min);

T_x is the analysis period (min).

The maximum E_p(max.) and minimum E_p(min.) percentage variations within an observation window of duration period P min are given by:

$$E_p(\text{max.}) = \underset{j=1}{\overset{m}{\text{MAX}}} \left[\frac{S}{P} \times \sum_{i=j}^{j+\frac{P}{S}-1} 100 \times \left(\frac{Q_i - r}{r} \right) \right] (\%) \tag{3}$$

$$E_p(\text{min.}) = \operatorname*{MIN}_{j=1}^{m}\left[\frac{S}{P} \times \sum_{i=j}^{j+\frac{P}{S}-1} 100 \times \left(\frac{Q_i - r}{r}\right)\right](\%) \qquad (4)$$

where

$$Q_i = \frac{60\,(W_i - W_{i-1})}{Sd}(\text{ml/h})$$

W_i is the i^{th} mass sample from the analysis period T_x (g) (corrected for evaporative loss);

r is the rate (ml/h);

S is the sample interval (min);

P is the observation window duration (min);

d is the density of water (0,998 g/ml at 20 °C).

The trumpet algorithm represents a method for obtaining a percentage error from the variations in flow rate, rather than an error on the measurements taken. It separates the measurement error and the pump error, allowing an objective and comparable analysis across various pumps at different flow rates.

The other aspect of the standard related to volumetric flow rates breaks them into two categories: minimum and intermediate flow rates. These are not actually set by the standard, instead they are set by the manufacturer (us), such that the minimum accuracy requirements for both rates can be met. In addition, they outline testing apparatus, which is detailed in the long-cycle method for verification below.

Mechanical Drawings of Pump Chassis

Connection Diagrams for System Setup

The following diagrams show the connections for our two custom circuits, the oscillator and driver. The RPi0 pins referenced in the diagrams follow the pin ordering labelled directly on the prototype. All connections are made using wires with male header pins on both ends.

Figure 10: Colpitts Oscillator PCB with labelled connections

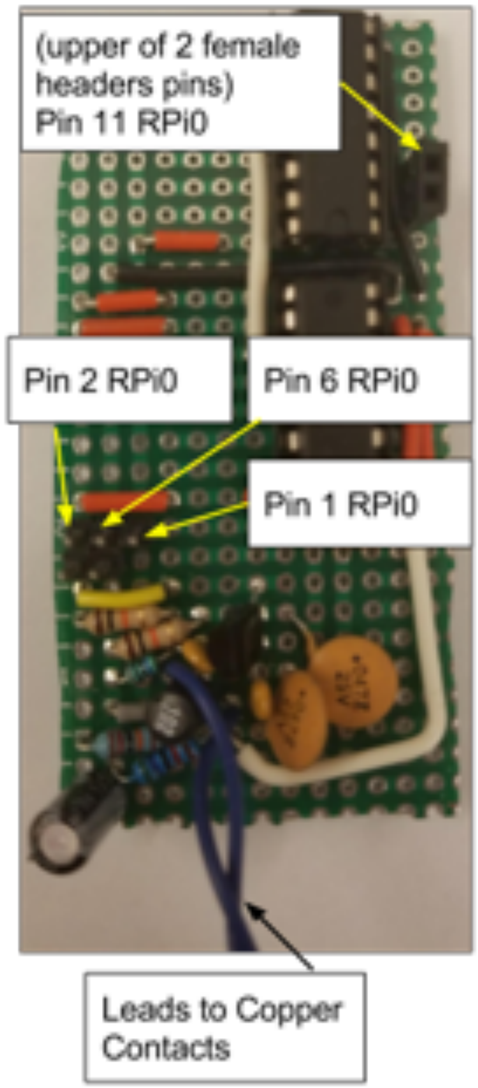
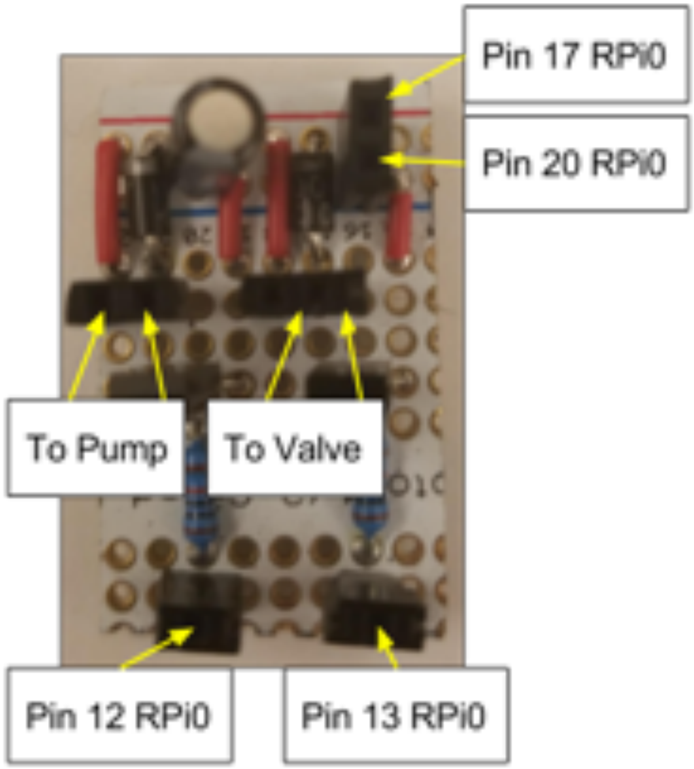


Figure 11: Motor Driver PCB with labelled connections



Software Information for Project

All code for the syringe pump developed during our project can be found at: https://github.com/akshivbansal/enph479

In order to run the Raspberry Pi Zero as headless (without using a keyboard and display, and just connecting via SSH), we followed the reference: https://davidmaitland.me/2015/12/raspberry-pi-zero-headless-setup/