PUMP DESIGN FOR A PORTABLE RENAL REPLACEMENT SYSTEM

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PUMP DESIGN FOR A PORTABLE RENAL REPLACEMENT SYSTEM

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To my family.

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LIST OF SYMBOLS AND ABBREVIATIONS

BUN	Blood Urea Nitrogen	mg/dL
DC motor	an electric motor that runs on direct current (DC) electricity	
ESRD	End Stage Renal Disease	
PD Pump	Positive Displacement Pump	
K_0	overall mass transfer coefficient	cm/min
K_0A	dialyzer properties	mL/min
RPM	Rotation Per Minute	/min

SUMMARY

Most patients diagnosed with End Stage Renal Disease (ESRD) undergo hemodialysis. Traditional hemodialysis treatment requires patients spending three to five hours every other day while yielding the high waste level accumulated between treatments. These limitations in the current technology have spurred the development of a portable renal replacement system. The portable system will not only free the patients from visiting the clinic but also allow more frequent treatment that will lead to lower average waste level.

To realize a portable system, the size and weight of hemodialysis system components should be reduced. This work analyzes the working principle of the pump and proposes a DC-motor and cam driven finger pump design. In addition, an analytical pump model is created for the design space exploreation. *In vitro* experiment conducted using the pump measured Creatinine levels over time, and the results validitate the design for the portable renal replacement system.

The proposed pump design is smaller than 188 cm^3 and consumes less than 3W while providing a flow rate of more than 100ml/min for both blood and dialysate flows. The smallest pump of a portable renal replacement system in the literature uses check valves, which considerably increase the overall manufacturing cost and possibility of clogging. Compared to that pump, the proposed pump design achieved reduction in size by 40% and savings in energy consumption by 65% with the removal of valves. This simple and reliable design substantially enables development of a portable renal replacement system.

CHAPTER ONE

MOTIVATION FOR A PUMP DESIGN FOR A PORTABLE RENAL REPLACEMENT SYSTEM

Only recently have researchers begun to develop a portable renal replacement system. [1] [2] Although Gura and Rambond developed a wearable kidney system in 2005 [3], the system has not yet been commercialized, and it still requires improvement. This chapter briefly explains the motivation for this thesis and introduces the research questions and hypothesis to be answered followed by the validation and verification strategies.

1.1 Portable Renal Replacement Systems and Pumps

End stage renal failure (ESRD) is a disease afflicting hundreds of thousands of patients worldwide. Most patients diagnosed with ESRD undergo hemodialysis. Traditional hemodialysis requires patients to visit a clinic three times a week for three to five hour long treatment sessions. This results in the patients disrupting their normal everyday life and restricting extended travel. In addition, because of the treatment schedule thatrequires visits to a clinic every other day, the high waste level accumulated for two days often makes patients experience dizziness or become lethargic. [4] These limitations in the current technology have spurred the development of a portable renal replacement system. The portable system will not only free the patients from visiting the clinic but also allow more frequent treatments that will lead to lower average waste level.

However, only a few researches have been conducting research on a portable renal replacement system. Gura et al. [3] developed a prototype wearable renal replacement device. Their device did not introduce significant advances in hemodialysis technology itself, but they attempted to transform the hemodialysis experience. The master's thesis of

Olson [5]seeked to improve hemodialysis in two ways: by proposing a new renal replacement therapy that does not rely on traditional hemodialysis components and by investigating the feasibility of adapting current hemodialysis practices to a portable format. As a result of the investigation, Olson et al. [6]created a model of dialysis treatment that can be used to analyze different treatment regimens. This work suggested developing a new portable pump since the size of the pump has the most room for improvement among the mechanical components of a hemodialysis system.

Current hemodialysis systems use peristaltic pumps because they can maintain a sterile flow path for the blood. However, they are inefficient and too large and heavy for portable applications. To reduce the size of the pump required for a portable system, Gura et al. [7] developed a wearable kidney pump that has a mass of 400 grams. Although this pump is very small compared to the peristaltic pump used in the current hemodialysis system, it could still be smaller and lighter. And the motor in the pump generates noise. Furthermore, the tubes used for the pump contain check valves, whose manufacturing costs will be far more than those of manufacturing tubes with no valves. In addition, those valves increase the possibility of clogging. To overcome these drawbacks, the development of a novel pump is required.

1.2 Research Questions and Hypothesis

The goal of this thesis is to design a novel pump for a portable renal replacement system. This thesis seeks to answer the following research questions:

Research Question 1:

Can a pump that is more compact, lighter, and more energy efficient than *currently existing pumps for the portable renal replacement system be designed with lower manufacturing cost, less noise, and less possibility of clogging?*

Hypothesis 1:

By adopting a different working principle and a different actuator, we can design a pump that is more compact, lighter, more energy efficient than currently existing pumps for the portable renal replacement system with lower manufacturing cost, less noise, and less possibility of clogging.

One of the pumps that best fit the intended application is the pump Gura et al. developed. The pump requires two check valves to prevent backflow, which would considerably increase the manufacturing cost of the tubes compared to the tubes with no valves. The price of tubes is a big concern because the pump requires daily replacement of tubes. In addition, the valves increase the possibility of clogging. If a different working principle could prevent backflow in a different way replacing the role of valves, the manufacturing cost and the possibility of clogging will decrease. In addition, the pump uses a DC-motor as an actuator. The DC-motor comprises half of the volume of the pump. If the volume and weight that the motor currently accounts for decreases, a pump would be more compact and lighter. Furthermore, the motor makes noise, which makes the treatment even more uncomfortable.

Research Question 2:

Can an analytical model that embeds relationships between limited number of design parameters (which includes design requirements, design specifications, and *characteristics of an actuator) be reliably used to explore the design space under a predefined experimental setting?*

Hypothesis 2:

An analytical model that embeds relationships between limited number of design parameters can be reliably used to explore the design space by introducing a lumped constant that accounts for the predefined experimental setting.

Numerous design options are available for generating the same flow rate. One could use a big tube and run the pump slowly or use a small tube and run it fast. However, those two options will have different sizes and consume different amounts of energy. To find the best option among those, a designer should know how the whole design will be affected by changing each design parameter such as the size of the tube or target flow rate. A pump model that embeds all those relationships between parameters will be able to help the design process by providing the expected pump specification for different combinations of design parameters without actually building a pump and running experiments.

This thesis will present an analytical model that embeds relationship between limited number of design parameters. Those parameters will include design requirements, design specifications, and characteristics of an actuator. However, experimental setting information such as fluid property, pump head, friction loss, and backflow will not be considered individually but will be represented by a lumped constant. Once the lumped constant for the given setting is found from an experiment, the model can be reliably used to explore the design space. The pump model can be used not only for the pump design of a portable renal replacement system but for any pump that has the same working principle as long as a predefined experimental setting is known.

Table 1-1. Organization of research questions.

Research Question	Corresponding Chapter
Question 1	Chapter 3, 4
Question 2	Chapter 5, 8

1.3 Validation and Verification Strategy

The primary aspects of this thesis in need of verification are the pump model created in Chapter 5 and the pump designed in Chapter 6 for the portable renal replacement system. Because the model predicts the performance of the pump in a given setting, its validity is important to generalize the model for this particular design so that it can be used for predicting the performance of the pump in different settings without actually making the working prototypes every time the setting changes.

In Chapter 7, experiments carried out with water compare the results of the model with the experiment results. Then, *in vitro* experiments conducted at the Georgia Institute of Technology in Dr. David Ku's laboratory compare the results of the model with those of experiments (flow rates of the blood and the dialysate). Finally, *in vitro* experiments conducted measuring Creatinine levels over time verify the performance of the designed pump for a portable renal replacement system by comparing it with a simulation result [5].

1.4 Closure

This thesis investigates a pump design for a portable renal replacement system from a systematic engineering design perspective and creates an analytical pump model for a certain kind of pump to facilitate the systematic design process and explore the design space. Chapter 1 provides the motivation and the context of this work. Chapter 2 provides the background information. Chapter 3 provides the conceptual design of a pump by following the steps of the Pahl and Beitz method. Chapter 4 investigates finger pumps and provides general understanding of the finger pump characteristics. Chapter 5 documents the creation of a pump model using Simulink and investigates the behavior of design paramters. Chapter 6 documents the detail design of the pump in Solidworks. Chapter 7 verifies the pump model with experimental data and the pump with *in vitro* experiment data. Chapter 8 suggests the optimum pump design for a portable renal replacement system. Chapter 9 concludes this thesis by providing an analysis of the provided work, acknowledgements, and suggestions for future work.

CHAPTER TWO

PRESENT WORK IN PORTABLE RENAL REPLACEMENT SYSTEMS

Several renal replacement therapies are available for ESRD patients. Among those, hemodialysis treatment is the most common one and 325,000 patients are receiving the treatment only in the U.S. However, because of many drawbacks of the treatment, a few researchers have begun to develop a portable version of the current hemodialysis system. This chapter will provide the background information about current hemodialysis and will present development of portable renal replacement systems including pumps for those systems.

2.1 Hemodialysis System

According to Olson [5], hemodialysis mimics the general behavioral strategy of the kidney in filtering the blood of waste as shown in Figure 2-1. During the treatment, solutes are filtered by diffusion across a semipermeable membrane.

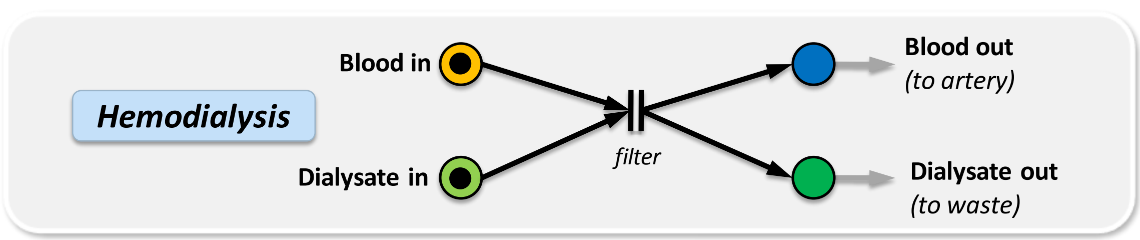


Figure 2-1. Functional performance of hemodialysis (Olson [5])

The dialyzer (labeled *filter* in Figure 2-1), which contains the semipermeable membrane, is the primary component of modern hemodialysis. In the dialyzer, to create a concentration gradient across the membrane, the blood flows along one side of the membrane countercurrent to the dialysate flow.

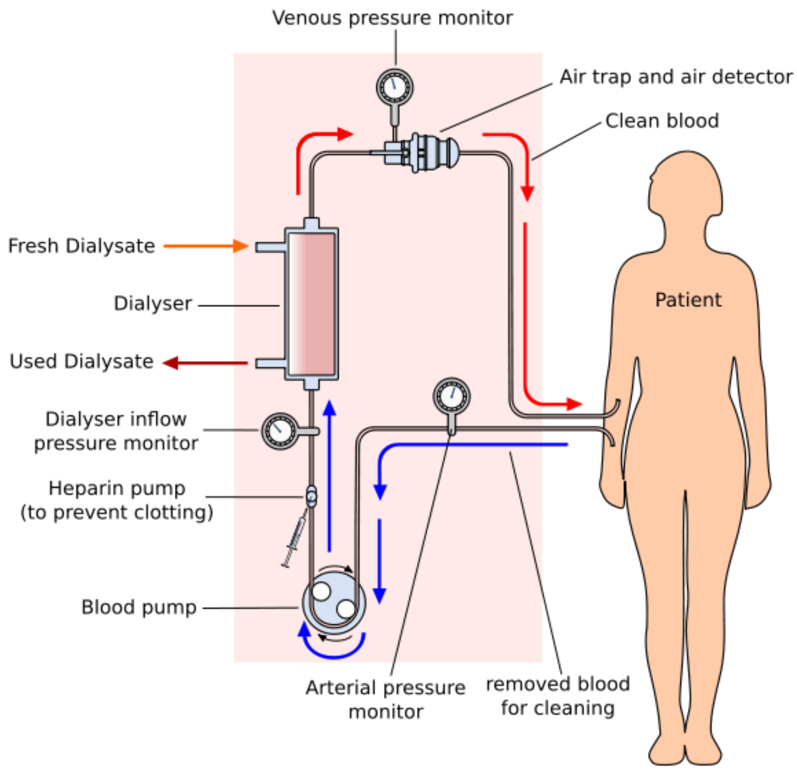


Figure 2-2. A hemodialysis circuit (Yassine Mrabet [8])

In a typical hemodialysis system (Figure 2-2), the blood leaves a patient's body through an implanted port and flows through medical tubing. It passes through a pump to keep the pressure and enters the blood compartment of the dialyzer. Then, the cleaned blood returns to the body at approximately the same pressure at which it was drawn from the body. In the process, heparin is added to prevent blood coagulation. To minimize infections and complications, the loop is kept simple and clean as much as possible. The dialysate loop is similar except that the dialysate is drawn from a reservoir before it is pumped through a dialyzer and disposed after the dialysate has run through the dialyzer counter to the flow of the blood.

During the hemodialysis treatment, small molecules diffuse across the membrane to areas of lower concentration. The dialysate contains no solutes to be filtered, while it keeps the same concentrations of those solutes to be kept in the blood. However, hemodialysis treatment is not very efficient at removing harmful middle molecular weight solutes from the blood. In addition, it requires patients to visit a clinic three times a week for three to five hour long treatment sessions. This schedule not only ruins patients' normal daily life, but also is not healthy for patients because harmful solutes are built up for more than two days before the next treatment resulting in high Blood Urea Nitrogen (BUN), a measure of urea content in the blood. Patients with high BUN levels experience symptoms of fatigue and poor fortitude.

2.2 Portable Renal Replacement Systems

2.2.1 Wearable Kidney Patent

Gura et al. [3] developed a prototype wearable renal replacement device. Their device did not introduce significant advances in hemodialysis technology itself, but their attempt to transform the hemodialysis experience was noteworthy. In order to overcome issues surrounding the large amounts (more than 100L per treatment) of dialysate used during a hemodialysis dose, the creators employed modified REDY sorbent cartridges to regenerate dialysate. After passing through the dialyzer and becoming dirty, the dialysate passes through a sorbent cartridge where the wastes in the dialysate are absorbed using some chemicals (such as activated carbon, urease, zirconium phosphate, and hydrous zirconium oxide). Then, the cleaned dialysate returns to the reservoir instead of being discarded. According to their 2007 patent filing of their best embodiments [7], they claim to use less than 1 L of dialysate. In addition, preliminary tests of the device on six pigs had promising results without any complications.

2.2.2 Portable Renal Replacement System

Olson [5]sought to improve hemodialysis in two ways: by proposing a new renal replacement therapy that does not rely on traditional hemodialysis components and investigating the feasibility of adapting current hemodialysis practices to a portable format.

While an alternative renal replacement therapy may be the best solution to today's dialysis problems such as low efficiency in removing middle size molecules, this work further focused on reducing hemodialysis to a portable format through systematic engineering design to improve patients' life quality by reducing the time they have to spend at clinic to get the hemodialysis treatment. As a result of the work, Olson et al. [6] created a model of dialysis treatment that can be used to analyze different treatment regimens and verified through *in vitro* experiments carried out with porcine blood and published human hemodialysis data. After model verification, they generated hemodialysis concepts that allow for maximum portability under different patient conditions. Although Olson [5] primarily focused on reducing the amount of the dialysate, the broad investigation on the system included information about a pump for the portable system. It showed the poor efficiency of the commercial pumps and suggested to develop a positive displacement pump for the portable system.

2.3 Pumps for Hemodialysis Systems

In the case of hemodialysis system design, maintaining a sterile flow path for the blood is the most important issue. Any pump used in the system must not introduce harmful agents (such as bacteria or viruses) to the blood or dialysate to maintain sterility. The peristaltic pump can easily provide this safeguard and performs adequately with the flow characteristics of the system.

Current hemodialysis systems use peristaltic roller pumps because they can maintain a sterile flow path for the blood. Peristaltic pumps are one kind of a rotary positive displacement pump. Moving rollers in a peristaltic pump squeeze a flexible tube forcing the fluid inside to move in one direction. Because the flexible tubing is the only part exposed to the fluid, the pump can easily maintain sterility. The life of the pump is adequate because only the flexible tubing requires replacement at regular intervals, and they are noted for their superior metering abilities. However, according to Olson [5], current peristaltic pumps show very poor efficiency in the range of flow rates associated with a portable hemodialysis system–well below 0.30%. Most of energy loss comes from squeezing very stiff tube. However, using a compliant tube is not an option because the tube is required to keep its own shape while it is bent in the pump. In addition, the pump is very noisy due to the motor inside the pump. Moreover, the pump is bigger than 11.5 × 11.5 × 30 cm and weighs more than 15 N even without a control unit. Considering that two pumps, one for the blood flow and one for the dialysate flow, are required for a hemodialysis system, this pump cannot be used for the portable application.

To overcome these drawbacks of current peristaltic pumps, Gura et al. have developed a specialized pump for their prototype wearable artificial kidney, which has not yet been commercialized. For their device, they have developed a specialized dual channel pulsatile pump that is actuated by a motor controlled by microcontrollers. The pump, 9.7 × 7.1 × 4.6 cm big and with mass of 400 grams, requires less than 10 W of power for operation. Although this pump is very small compared to the peristaltic pump used in the current hemodialysis system, it could still be smaller and lighter. In addition, the motor noise should be removed for the portable application. Furthermore, the tubes used for the pump contain check valves, whose manufacturing costs will be far more than those of manufacturing tubes with no valves. The price of tubes is a big concern because the portable pump will require the daily replacement of tubes. In addition, those valves increase the possibility of clogging. To overcome these drawbacks, the development of a novel pump that achieves the target flow rate with reduced size and weight of the pump, cost, noise, and possibility of clogging and increased energy efficiency is required.

2.4 Conclusion

Current hemodialysis treatment has many drawbacks such as low efficiency in removing middle sized molecules, large amount of time that the patients have to spend at a clinic, and the high BUN level of patients that the less frequent treatment brings. To overcome these drawbacks, current hemodialysis should be improved. The portable form of the system will reduce the amout of the time that the patients have to spend at a clinic by allowing them to do the treatment at home. In addition, the convenient treatment will allow more frequent treatment sessions which will keep the BUN level of the patient lower, compared to current hemodialysis treatment conducted every other day. The portable system requires smaller components than the current hemodialysis system, and the pump has the most room for improvement, among the mechanical components of the system.

This thesis will attempt to design a novel pump for a portable renal replacement system while removing the drawbacks of current hemodialysis treatment and the patented wearable kidney pumps.

CHAPTER THREE

CONCEPTUAL DESIGN OF A PUMP FOR A PORTABLE RENAL REPLACEMENT SYSTEM

This chapter is devoted to the conceptual design of a pump. The Pahl and Beitz method, from the task clarification to the embodiment design phase, was used to design a pump that is more compact, lighter, and more energy efficient than existing pumps with lower manufacturing cost, noise, and less possibility of clogging. This chapter seeks to answer the first research question introduced in 1.2 and concludes by proposing a possible layout of a pump design including possible actuators of the pump.

3.1 Planning and Clarification of Task

3.1.1 Task Clarification

This section clarifies the objective behind designing a pump for the portable renal replacement system. The given problem statement is as follows.

Problem statement:

To provide an improved portable renal replacement system, a novel pump is required.

As introduced in Chapter 1, the size of the system is one of the many shortcomings of the current hemodialysis system. A system that is small enough to be portable will improve the quality of life, granting newfound mobility to patients that are currently tethered to a machine several hours every other day. For a smaller portable system, smaller and lighter components would be necessary. Although the pump that Gura et al. [3] developed is smaller than the pump used in the current hemodialysis system, it requires check valves for the tube, resulting in increased manufacturing costs of the tube and possibility of clogging.

The increased cost leads to increased operating costs because the tube has to be replaced after every treatment. In addition, the pump uses a motor that comprises half of the volume of the pump and yields uncomfortable noise. Based on the problem statement, the problem is abstracted as follows.

Clarified task:

How can we design a pump that is more compact, lighter, and more energy efficient than currently existing pumps for the portable renal replacement system with lower manufacturing cost, noise, and less possibility of clogging?

Using the abstracted ideas proposed to minimize the size, weight, operating cost, noise, and possibility of clogging of the pump, this thesis will propose a number of promising products, descriptions of which follow.

Product proposal:

- *A pump that does not have a motor:* Greater ease of portability and mobility require smaller and lighter components. Because a motor is a sizable component in a pump, the removal of a motor could be a solution. In addition, the removal would also remove the motor noise. The use of an artificial muscle, shape memory alloy, or ultrasonic motor might be an answer.
- A pump that works according to a different working principle from the ones used in the current hemodialysis system or the wearable kidney patent: The pump of the wearable kidney patent [7] requires check valves, the components that block backflow in the tube. However, by exploring other working principles, we could remove the valves. The peristaltic pump used in the current hemodialysis system does not require valves because the moving rollers can completely block the

backflow without any valves. The application of a different working principle might allow the removal of valves from the tube.

- *A pump that requires a smaller battery:* Considering that the system will run from a portable energy source such as a battery, a more energy efficient pump would better serve the portable system because it would require a smaller and lighter battery than a less efficient pump. The size reduction of the required battery will lead to a size reduction of the entire pumping unit.

3.1.2 Requirements List

The requirements list will delineate the demands and wishes from the system that shall be designed. Demands are those expectations that a system will have to fulfill. Wishes are those expectations in the system, whose fulfillment is not a must, but which are preferred. Table 3-1 provide the requirements for the pump design.

Table 3-1. Requirements list for pump designs for the portable renal replacement system

D/W	Requirements	Responsible: Jane Kang
	Geometry:	
W	**1. Dimensions of the blood and the dialysate pump $\leq$ 200 cm^3** : The wearable artificial kidney patent suggested a pump of 317 cm^3. This thesis attempts to reduce the volume of the pump to 60%.	
W	**2. Minimal number of moving parts** : A number of moving parts is a factor that affects vibration, manufacturing cost, and durability. A low number of moving parts yields a better pump for the portable system.	

Table 3-1 continued

D	**3. Minimal contact area between the blood and moving parts and a minimal duration time** : The contact area between the blood and moving parts and duration time are factors that determine the rate of hemolysis, which is the burst of red blood cells.
D	**4. No infection (No contact between the blood inside and outside of the pump)** : Since absolutely no infection must be allowed to occur, outside the pump should not be connected with the blood inside.
Kinematics:	
D	**5. Blood flow rate $\geq$ 80 ml/min**
D	**6. Dialysate flow rate $\geq$ 91 ml/min** : These target flow rates are calculated from the MATLAB simulation [5] assuming a situation in which a patient uses the portable system every day.
Forces:	
W	**7. The mass of the blood and the dialysate pump $\leq$ 0.2 Kg** : The wearable artificial kidney patent suggested a pump (that pumps two fluids at once) of 400 grams. The desired mass is about 200 grams.
D	**8. No vibration**
D	**9. Less blood cell damage (Maximum shear stress in the blood flow $\leq$ 1000 Pa)** : Blood cell damage can be a serious problem if the damage exceeds a critical value. To reduce blood cell damage, the maximum shear stress in the blood flow needs to be below 1000 Pa. [9] [10]

Table 3-1 continued

	Energy:
W	**10. Energy consumption (for all pumps together) $\leq$ 5 W** : As mentioned earlier, energy consumption is an important factor that determines the size of the pump. The wearable artificial kidney patent pump consumes less than 10W.
D	**11. Temperature $\leq$ 37.5°C**
	Material:
D	**12. No harmful reaction on the part having contact with the blood** : Biocompatibility of the material
W	**13. Less friction at the contact point**
	Signals:
W	**14. Controllability of flow rates**
W	**15. Blood pressure monitor**
D	**16. Detection of leaks**
	Safety:
D	**17. Automatically stops when disconnected or leaked**
	Ergonomics:
W	**18. Easy for patients to understand and operate**
W	**19. Appearance not distracting when worn**
D	**20. Tied to the right place and stable**

Table 3-1 continued

	Operation:
W	**21. No distracting noise** : Both the peristaltic roller pump that is currently used and the patent pump use a motor to actuate the motion, which generates noise.
	Maintenance:
D	**22. Minimal blood clogging** : The patent pump contains two valves that increase the possibility of blood clogging.
W	**23. Less need to visit the clinic (Easy to keep the pump clean at home)**
	Costs:
D	24. **Lower cost than current hemodialysis system**

3.2 Conceptual Design

3.2.1 Identify Essential Problems (Abstract Requirements List)

The conceptual design phase starts by identifying the root of the problem, which should be stated in neutral terms so that it does not define the solution. Using the problem definition and the requirements list, the root of the problem must be identified. Then, it will be expanded using abstraction so that the generalized essentials of the problem, not the specifics of the problem, are identified. The abstraction is conducted as follows.

First, personal preferences are eliminated. Then, requirements that have no direct bearing on the function and the essential constraints are omitted. The results are as follows:

- Minimum number of moving parts

- Minimum contact area between the blood and moving parts and the duration time
- No infection (No connection between the blood inside and outside of the pump)
- Minimal blood clogging
- Energy consumption (for all pumps together) $\leq$ 5W

Then, quantitative data are transformed into qualitative data and reduced to essential statements. The results are as follows:

- Reduce the number of moving parts, contact areas between the blood and moving parts, the duration time of the contact, and possibility of blood clogging.
- No connection between outside of the pump and the blood inside
- Minimize energy consumption.

Then, the results of the previous step can be generalized as far as it is purposeful. The results are as follows:

- Reduce the number of moving parts, the contact area between the blood and moving parts, and the duration time of the contact.
- Disposable case is required for keeping the blood clean. The pump should operate outside the tubing.
- Reduce the required battery size and factors that cause blood clogging.

Finally, the problem is formulated in solution-neutral terms. The results are as follows:

- Design a pump that uses a disposable case for blood access with reduced required battery size and blood clogging. At the same time, the number of moving parts, contact areas between the blood and moving parts, and the duration time of the contact should be minimized.

Following the abstraction process, the root of the problem became evident.

3.2.2 Establish Function Structures

Figure 3-1 shows the function structure of the portable renal replacement system. In the system, the dotted red box on the left is the blood pump, and the dotted blue box on the right is the dialysate pump.

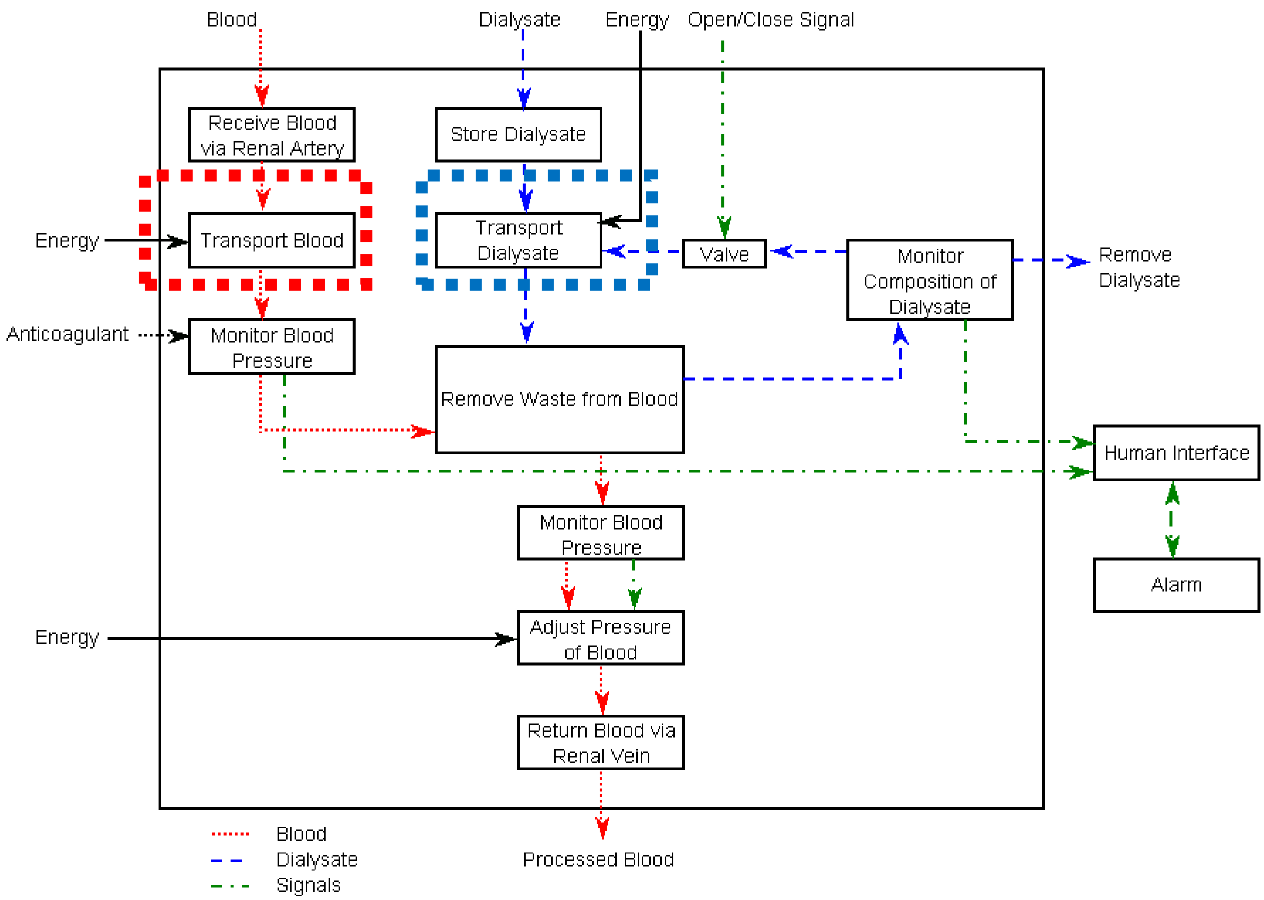


Figure 3-1. The flow path of energy, fluids, and information in a theoretical portable renal replacement system (Olson [5])

Figure 3-2 shows the function structure of the pump. The reason that the pump has a storage function is because when the pump is not operating, the blood or dialysate stays in the pump, which is a storing, not transporting function.

Figure 3-2. Function structure of the pump

Having the "storage function" as a sub-function is also part of the abstraction. The abstraction phase concluded that the pump requires a separate case for the blood that can be disposed after each use. For example, a flexible tube will be the disposable case for the blood or the dialysate responsible for the "storage function." Thus, the transportation of the dialysate or the blood should be conducted outside the flexible case.

3.2.3 Search for Working Principles and Working Structures

3.2.3.1 Literature Search and Analysis

To find the working principles of existing pumps, the conventional methods of a literature search and the analysis of existing pumps were used. Since the pump has existed for so long, the literature search is the best option for the start. From the pump search and study, a pump tree is built, and it is shown in Figure 3-3. A bigger version of Figure 3-3 is available in the appendix.

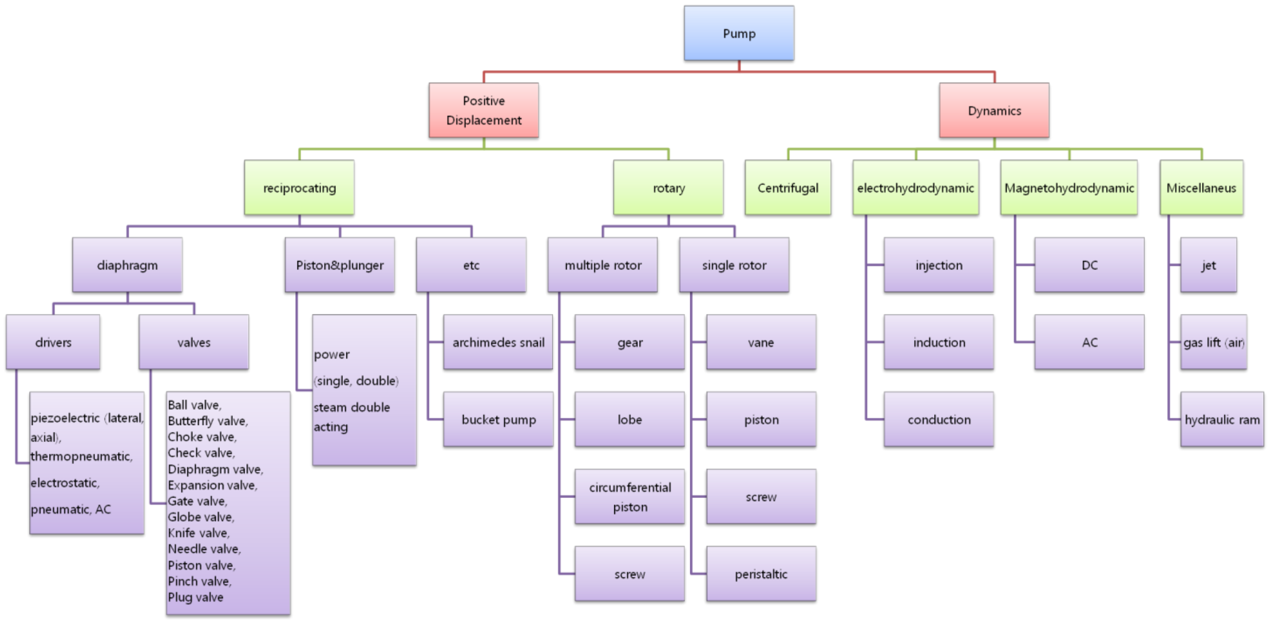


Figure 3-3. Pump tree

The pump tree divides pumps into positive displacement (PD) pumps and dynamic pumps.

Figure 3-4. Impeller pump

Figure 3-4 shows an impeller pump, a common dynamic pump. A dynamic pump causes the fluid to move from an inlet to an outlet under its own momentum. In the figure, the impeller in the center will generate centrifugal force to move fluid from the inlet in the center to the outlet shown in the bottom of the picture. Other than using centrifugal force, a dynamic pump could use elctrohydrodynamic force, magenetohydrodynamic force, etc. None of the dynamic pumps can have a separate case such as flexible tubing because all the dynamic pumps need a connection between the outside and the inside of the case of fluid such as a shaft that provides centrifugal force. An electro hydrodynamic pump might not require such a connection, but it is still not available because of the need to add something to induce electric power, which is not recommended for this application. Allowing direct contact with the fluid and the pump actuation parts will lead to a possible infection and blood clotting problem.

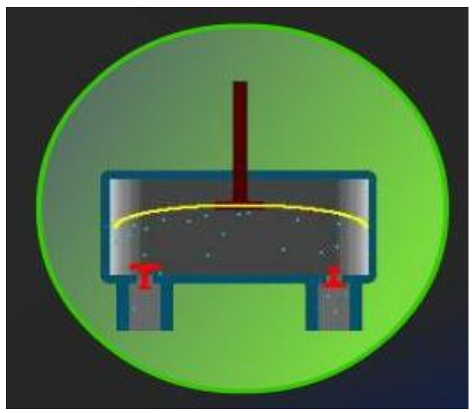


Figure 3-5. Diaphragm pump

Figure 3-5 shows a reciprocating type of PD pump, diaphragm pump. A PD pump causes a fluid to move by trapping a fixed amount of it and forcing (displacing) that trapped volume into the discharge pipe. For the diaphragm pump shown above, two check valves allow the fluid to move only in one direction (left to right) as diaphragm moves up and down repeatedly.

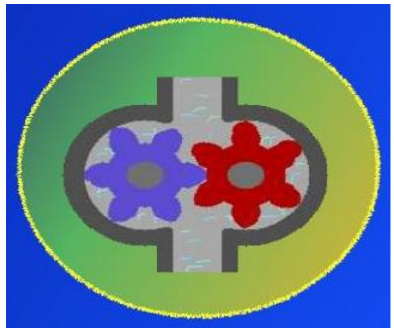


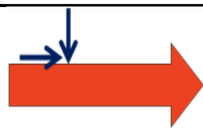
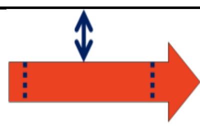
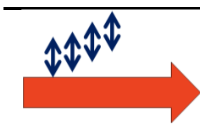
Figure 3-6. Gear pump

Figure 3-6 shows a rotary type of PD pump, gear pump. In this case, the rotating gears trap a fluid causing it to move in one direction. The PD pump is applicable in the use of a separate case from the pump. (For the above pumps, imagain a separate case holds the fluid and reciprocation or rotation is applied right outside the flexible case.) Thus, only the working principles of PD pumps will be analyzed.

3.2.3.2 Working Principle of Positive Displacement Pump

Two main kinds of forces push the fluid in PD pumps. One is the transient force, which is applied in the transient direction of the flow, and the other is the normal force, which is applied to the normal direction of the flow. All PD pumps operate by combining these two forces. The working principles of the PD pump are introduced in Table 3-2.

Table 3-2. PD pump working principle

Applied forces		Examples	How to provide force	Pros and Cons
Transient Force		Piston pump, Rotary pumps	Piston, Gear, Lobe	Con: Direct contact between the fluid and pump (no separate case)
Transient + Normal		Peristaltic roller pump, Gear pump	Gear, Cam, Screw	Pro: No direct contact Con: Friction loss when roller is not used.
Normal Force		Diaphragm pump, Piston pump	Cam, Artificial muscle, Electromagnetic, Shape memory alloy wire, Ultrasonic linear motor	Pro: No direct contact Con: Requires valves
		Finger pump		Pro: No direct contact

The first type uses only the transient force. However, this type of pump applies force inside the case of the fluid. Thus, the fluid and the pump have direct contact could cause infection and blood clotting. As a result, this type of pump is excluded as a candidate for a pump design.

The second type uses both the transient force and the normal force applied outside the tubing. Examples of this type of a pump is the peristaltic roller pump, which is used for the current hemodialysis treatment, and gear pump shown in Figure 3-6. However, the gear pump involves friction loss because of the transient force applied outside the tubing. The friction loss does not exist for the peristaltic roller pump because of the roller compressing the tube. Negligible amount of friction loss occurs inside the bearing of the roller. As mentioned before, the peristaltic roller pump involves huge energy loss that comes from squeezing very stiff tube. Because the tube should be bent and still be able to keep its own shape, compliant tubes cannot be used.

The third type uses only normal force. Because only normal force operates the pump, a fixed amount of fluid cannot be trapped without valves, which involves blood clogging and complexity in the manufacturing process.

The fourth type uses sequential normal forces. Because the normal force applied to the tube can block the fluid from the back flow, no valve is required.

From the analysis of the PD pump working principle, we were able to exclude the first type of the pump because it cannot be applied with a separate case for the fluid. However, with both pros and cons, the other three types could be applied in a separate case for the fluid. Lubrication can reduce friction loss. However, at this point, the most energy efficient and the cheapest combination that requires the smallest size cannot be determined.

3.2.4 Combine and Firm Up into Concept Variants

To come up with concept variants, the method that can provide the actuation of a pump and forces are combinded with the working principles based on Table 3-2. Figure 3-7 depicts the morphological matrix.

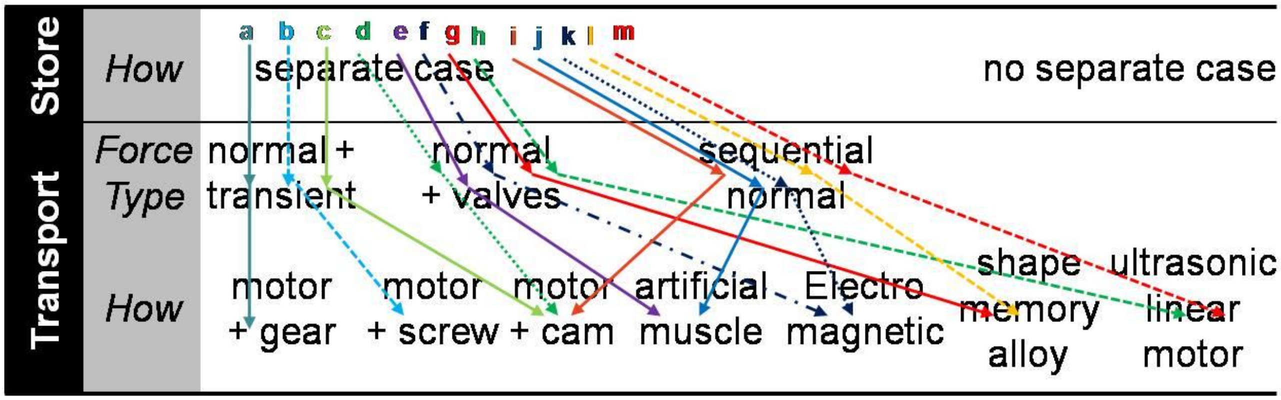


Figure 3-7. Morphological matrix with combined working principle

Because having a separate case is a must, the morphological matrix only has combinations with separate case. The separate case can be either a flexible tube or a balloon like membrane. Either of these will be applicable to any case, so it is not presented in the morphological matrix for simplification. Suggested combinations are explained in Table 3-3.

Table 3-3. Suggested combinations

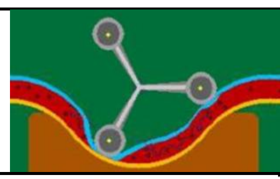
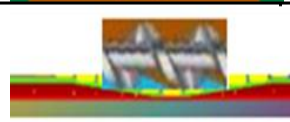
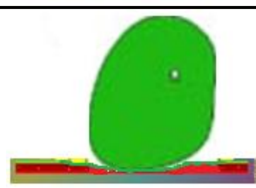
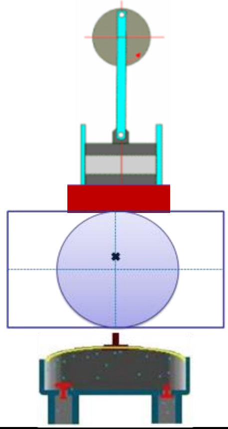
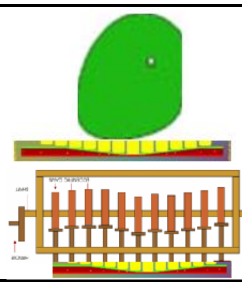
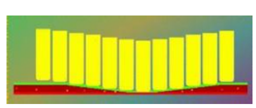
a. Normal+Transient & Motor+Gear		This is the peristaltic roller pump currently used for the hemodialysis treatment.
b. Normal+Transient & Motor+ Screw		A screw is acting outside the tube.
c. Normal+Transient & Motor+ Cam		A cam is acting outside the tube.

Table 3-3 continued

d. Normal+Valves & Motor+ Cam		A piston cam or a positive cam can be used to apply the normal force. It is the same working principle as the patent design.
e. Normal+Valves & Artificial Muscle		Artificial muscle (blue on the top and the bottom) is surrounding the tube (red in the middle), and squeezes the tube as nurses squeeze a blood pressure cuff. Check valves are required.
f. Normal+Valves & Electromagnet		The setting and working principle are the same as **e**. Instead of an artificial muscle, two plates are on the sides of the tube and attached and detached with electromagnet, shape memory alloy, or ultrasonic linear motor. Check valves are required.
g. Normal+Valves & Shape memory alloy		
h. Normal+Valves & Ultrasonic linear motor		
i. Sequencial Normal & Motor+Cam		The cam is acting on the plates that are placed outside the tube. Only normal force acts between the plates and the tube, applied in sequence. Friction loss between cam and plates still occurs, but it can be reduced by lubrication.
j. Sequencial Normal & Artificial Muscle		Many artificial muscles are surrounding the tube and squeeze in sequence.
k. Sequencial Normal & Electromagnet		The setting and working principle are the same as **e**. Instead of an artificial muscle, plates are on the sides of the tube and attached and detached in sequence with electromagnet, shape memory alloy, or ultrasonic linear motor.
l. Sequencial Normal & Shape memory alloy		
m. Sequencial Normal & Ultrasonic linear motor		

3.2.5 Evaluate Against Technical and Economic Criteria

Suggested combinations are evaluated based on the technical feasibility and their economic merit to allow a confident decision to be made about the most suitable principal concept variant.

Mainly four things were considered in the evaluation process. The patent design will be used for the comparison and assumptions in the evaluation process. Any design allows for improvement compared to the patent or matches the patent specifications without copying it will be considered as a candidate. To evaluate suggested designs, actuators and battery information was searched before the evaluation.

3.2.5.1 Considerations

- Possibility of clogging: Possibility of clogging can be evaluated depending on the number of check valves.
- Possible cost: Possible cost can be compared considering the cost of an actuator and the complexity of manufacturing process such as inclusion of valves.
- Volume and weight: A possible size and the weight of the pump can be estimated considering main parts included in the pump such as the volume and weight of actuator.
- Required size of battery: Required power can be estimated from the required power for actuators. Depending on the power requirement, the required battery can be determined, and this information can be used to estimate the weight of the battery that will affect the whole weight of the system.

3.2.5.2 Analysis of Wearable Artificial Kidney Patent

The pump of the patent that was developed by Gura and Rambond were similar to design d (Normal+Valves & Motor+ Cam). They used a 3W DC-micromotor that operates on 12V and consumes less than 10W. The patent did not specify what angular velocity (rpm) they used. However, they used spur gear heads 15/5, and from that information, it was deduced that the motor operates at 85 rpm with an output torque of 100mN.m. In addition, according to the drawing provided in the patent, the cam squeezes about 4 cm of the tube. Thus, 100 mN·m is assumed to be enough torque for squeezing 4 cm of the tube. The exact force cannot be calculated with the given information. Thus, it is assumed that 2.5 N/cm (10 N for 4 cm) is the force required to squeeze the tube. The motor and spur gear heads are 73mm long with ∅6. The entire size of the pump was 9.7×7.1×4.6 cm = 317 cm^3 and the mass is less than 400 g. The target flow rate was 100ml/min.

3.2.5.3 Actuator

3.2.5.3.1 DC-Motor

Many motor companies produce various kinds of DC micro motors. Among those, this thesis considered Faulhabor. DC micro motors of the company are very compact while producing large torque. The mass of the motor including gearhead varies from 4 grams to 1500 grams and the torque 0.25 Nm to 16Nm. The nominal voltage varies from 3V to 24V. According to requirements for actuators, the right DC micro motor can be chosen. The cost of a motor with gearhead is around $130.

3.2.5.3.2 *Artificial Muscle*

Artificial muscle could be a great candidate since the weight and the size are known to be very small compared to the force and displacement it generates. However, it is still in the stage of development and not many companies carry it.

3.2.5.3.3 *Electro Magnet*

Electro magnets are commercialized for pulling and pushing actuation as solenoids. Among many products, this thesis considered STA series tubular solenoids. The size of solenoids varies from 13.2mm to 44.4mm diameter and from 13.9mm to 119.7mm long. These solenoids can generate up to 43.61N force and stroke up to 63.5mm. A 20mm diameter and 40mm long pull tubular solenoid with mass of 83.6 grams and generates 4N force for 25% duty cycle with power consumption of 14W. The cost is around $80 each.

3.2.5.3.4 *Shape Memory Alloy*

This thesis particularly considered one kind of nickel-titanium shape memory alloy, Flexinol. Flexinol contracts when it is heated and goes back to the original shape when it is cooled down. According to the datasheet, the contraction range can be up to 4% of the total length, and the diameter of wires varies from 0.025mm to 0.375m. When Flexinol contracts, it generates up to 20N force. Since Flexinol is wire, the weight is negligible. The efficiency of Flexinol is less than 5% because most of the energy is dissipated as heat. The cost is around $5 per meter.

3.2.5.3.5 *Ultrasonic Linear Motor*

This thesis considered an ultrasonic linear micro motor, SQUIGGLE motors. Two models are developed and those have nanometer resolution. The speed can vary from 0.001mm/second to 7mm/second and it can generate up to 5N force. The mass of each motor is less than 2 grams and consumes less than 3W of energy. The cost is around $650 including a microcontroller for each SQUIGGLE motor.

3.2.5.4 Battery

To estimate the weight of the battery that will be required for each suggested design, many commercial batteries were condsidered. Table 3-4 shows each commercial battery's voltage, capacity, and size information.

Table 3-4. Commercial battery information

Voltage (V)	Ampere hour (Ah)	Watt hour (Wh)	mass (g)	volume (cm3)	density (g/cm3)
3.7	7	25.9	146	79.968	1.82573
12	1.8	21.6	85	N/A	N/A
3.7	2.6	9.62	46.5	17.09643	2.719866
1.2	3	3.6	62	N/A	N/A
3.7	2.6	9.62	68	N/A	N/A
3	0.565	1.695	5	N/A	N/A
3	0.9	2.7	17	7.270038	2.338365
3	1.4	4.2	20	N/A	N/A
9	1.2	10.8	40	N/A	N/A

Based on the capacity (Wh) and mass information provided, the relationship is shown as Figure 3-8. As the power requirement increases, the mass of the battery increases as well. Thus, reducing the power requirement naturally leads to the mass reduction. It also should be noted that the mass of battery is related to the capacity, not the voltage.

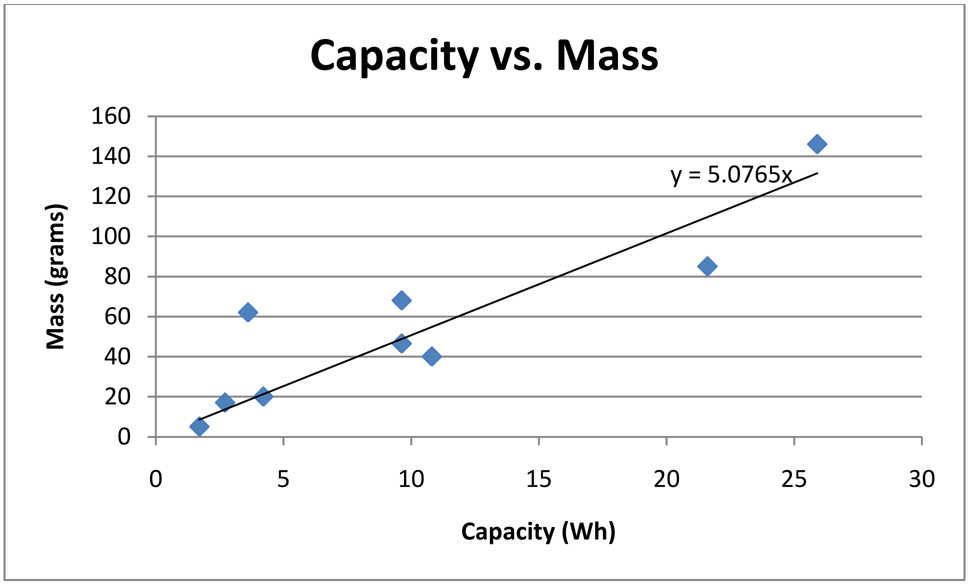


Figure 3-8. Battery capacity and mass relationship

From the figure, the relationship between commercial batteries' Watt-hour and mass is found as below.

$$\text{flow rate (ml/min)} = \text{flow amount of a finger stroke(ml)} \times \text{no. of finger strokes (/min)} \tag{3-1}$$

From the given density information, the density can be assumes as the average value, 2.3 g/cm^3, which gives below equations that relates the battery capacity to the battery size.

$$\text{Battery size constant } (\text{cm}^3/\text{Wh}) = \frac{5.0765\ (\text{g/Wh})}{2.3\ (\text{g/cm}^3)} = 2.2065\ (\text{cm}^3/\text{Wh}) \tag{3-2}$$

$$\text{Battery volume } (\text{cm}^3) = \text{battery size constant } (\text{cm}^3/\text{Wh}) \times \text{Watt hour (Wh)} \tag{3-3}$$

This relationship will be embedded in the pump model to convert power consumption information into the size information and estimate the total volume of the pump for a given setting.

3.2.5.5 Evaluation of Suggestions

Design **a** (Normal+Transient & Motor+Gear) in Table 3-3 is the same as the one used for the current hemodialysis treatment. As mentioned earlier, the loss between the roller and the tube is huge, and it is the same for design **b** (Normal+Transient & Motor+Screw) and **c** (Normal+Transient & Motor+Cam). In addition, design **a** requires the tube to be positioned as a circle. It requires certain stiffness of the tube, which leads to more force being required to compress the tube. When more force is required, bigger motors and more energy consumption follow.

For designs **b** (Normal+Transient & Motor+Screw), because of the volume of the screw, it is difficult to apply to our system. According to the ISO standards, 64 mm in diameter is required for 6 mm of pitch height. The volume of the screw and a motor that will drive the screw will easily be bigger than the pump of the patent.

Design **i** (Sequencial normal & Motor+Cam) operates with almost the same working principle as designs **a** (Normal+Transient & Motor+Gear) and **c** (Normal+Transient & Motor+Cam). However, it can be a better solution because little friction loss will occur between the tube and plates that compress the tube. Friction loss that occurs between plates and the cam can be minimized by lubrication. Lubrication was limited in designs **a** and **c** because the material used for the tube is a flexible one such as silicon. For this design, the use of metals and lubrication oils in the contact point of the cam and the plates can minimize friction loss. If no friction loss occurs in design **i**, the required torque will be the same as the patent design (design **d** (Normal+Valves & Motor+Cam)). Assuming that the friction coefficient between the cam and the plates is less than 0.1, less than 10% of additional torque will be required for design **i**. Thus, it will consume less than 10% more energy compared to design **d** and it will require no valves.

Using artificial muscle or shape memory alloy can be good in terms of the size and weight of the pump. Both the size and the weight could reduce significantly because those are very small compared to a motor. However, as mentioned before, not many companies carry artificial muscle, which makes it hard to use it for this pump design. In addition, using a new material, it may face hurdles to meeting regulations and standards when applied to the pump compared to others. It may also be costly to manufacture. The shape memory alloy can be a good actuator for the system in terms of cost because it is easy to get and inexpensive. However, the treatment of the heat that is generated during the actuation of the wire is a problem to be solved. An ultrasonic linear motor could be a good actuator for the system if only the force generated is enough because the size and weight of the design could be smaller than the patent design. However, the cost could be a big problem.

Design **f** (Normal+Valves & Electromagnet) uses an electromagnet to squeeze the tube. A possible actuator for this design is a tubular solenoid that can provide a stroke of 8 mm with a force of 10N, consumes 20Watts, and operates on 6.3 Volts. Its dimensions are ϕ26mm, 84mm with mass of 220 grams . This design could be smaller and lighter than the patent design.

Design **k** (Sequencial normal & Electromagnet) uses electromagnets to squeeze the tube in sequence. A possible actuator for this design would be a tubular solenoid that can provide a stroke of 8 mm with a force of 4N. The solenoid will consume 14 Watts, operate on 3.8 Volts. Its mass is about 100 grams with dimensions of ϕ20mm, 70mm. The design will require at least three tubular solenoids to operate in sequence. The size and the weight will be similar to the patent. It depends on the actual force required to squeeze the tube because smaller and lighter solenoids are available for smaller force requirement.

3.2.6 Principal Solution

Based on the evaluation process comparing the size, required power, and cost to the patent design, each combination was evaluated from XX(-2) to OO(2) for each consideration and added to show overall result. Table 3-5 shows the results in the next page.

As a result, designs **i**, **j**, **l**, and **m** remain as final concept design options. However, **j** may not be a good choice at this point because of the unavailability of the material (artificial muscle). Once the material can be procured, it would represent one of the most ideal solutions. Thus, this design will remain as an option for reference.

The result suggests that the use of sequential normal forces compressing a tube is the best option regardless of actuators (except for the solenoid that consumes large amount of power). This type of pump is commonly called a finger pump.

Table 3-5. Evaluation chart for the combinations in Table 3-3

Working Principle & Actuator	Clogging	Size	Power	Cost	Overall
Patent pump	X	=	=	=	**X**
a. Normal+Transient & Motor+Gear	O	X	X	=	**X**
b. Normal+Transient & Motor+ Screw	O	XX	XX	=	**XXX**
c. Normal+Transient & Motor+ Cam	O	X	XX	=	**XX**
d. Normal+Valves & Motor+ Cam	X	=	=	X	**XX**
e. Normal+Valves & Artificial Muscle	X	OO	=	X	**=**
f. Normal+Valves & Electromagnet	X	O	X	X	**XX**
g. Normal+Valves & Shape memory alloy	X	O	XX	=	**XX**
h. Normal+Valves & Ultrasonic linear motor	X	OO	O	XX	**=**
i. Sequencial Normal & Motor+Cam	O	=	=	=	**O**
j. Sequencial Normal & Artificial Muscle	O	OO	=	=	**OOO**
k. Sequencial Normal & Electromagnet	O	=	XX	X	**XX**
l. Sequencial Normal & Shape memory alloy	O	OO	XX	O	**OO**
m. Sequencial Normal & Ultrasonic linear motor	O	O	O	XX	**O**

3.3 Embodiment Design

In the conceptual design phase, finger pump is chosen as the best pump for the portable renal replacement system. This section will suggest two finger pump layouts for the final design concepts

3.3.1 Layout for Linear Actuators (Design l and m)

Figure 3-9 is the layout for design **l** (shape memory alloy).

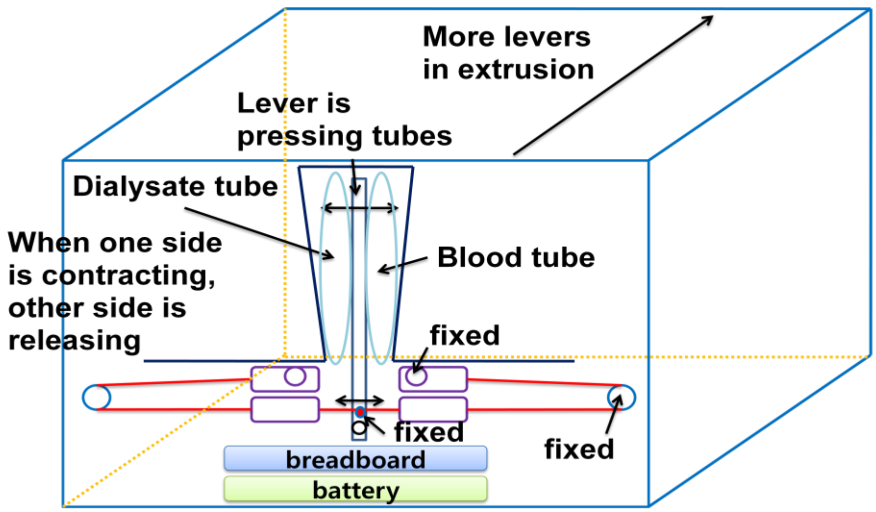


Figure 3-9. Layout of a finger pump design for Linear Actuators

The portable renal replacement system requires pumping of two different flows, the blood and dialysate. The suggested design allows pumping of two different flows with one pump. The fingers are positioned between the two tubes. Thus, when a finger compresses the tube containing dialysate, it opens the tube containing the blood to receive more fluid and vice versa. Hence, actuation is required in both directions. The sequential movement of fingers will make a typical finger pump that can force the fluid inside tubes to move in one direction.

The finger is fixed at a point, and one part of the finger is connected to the actuator while the other part is compressing tubes back and forth. The finger could be used as a lever that amplifies or reduces the displacement of the actuator depending on situations. The figure shows the situation where the actuation is amplified to compress the tube with a little actuation. (The fixed point is closer to where the actuator is connected than the other end compressing tubes.) In addition, when shape memory alloy is used, a pulley could be used to reduce the space that the long wire would take as long as it does not affect the functionality.

For design **m** that uses ultrasonic linear motor or any other linear actuators, all the other parts, such as the position of fingers and plates, can be kept the same except the actuator connection part shown with lines in Figure 3-9.

3.3.2 Layout for Motor-Cam Design (design i)

Figure 3-10 is the layout for motor cam design.

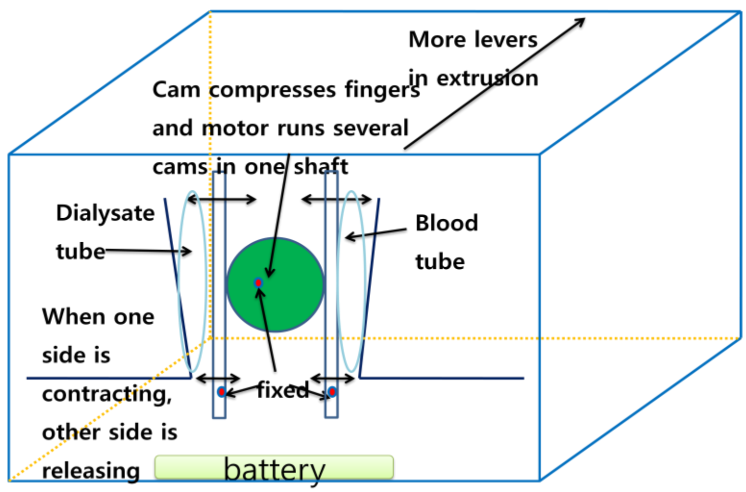


Figure 3-10. Layout of a finger pump design for motor cam design

The layout is very similar to the previous one. As before, the design allows pumping of two different flows with one pump. Cams are located in the middle of two rows of fingers to actuate fingers and compress each tube. The cams are mounted on the cam shaft with different orientations to provide sequential actuation. The fingers are mounted on two shafts as levers to make two rows of fingers, and tubes are positioned between fingers and flat plate of the main body.

3.4 Conclusion

In this chapter, a finger pump is selected as the final candidate, and layouts are suggested. A finger pump is expected to be more compact, lighter, and more energy efficient than currently existing pumps by removing the friction loss accompanied by currently used peristaltice pumps. In addition, the finger pump that requires no valves is expected to yield lower manufacturing cost and less possibility of clogging.

Motor-cam, Flexinol (shape memory alloy), and ultrasonic linear motor are left as possible candidates for an actuator for the finger pump. Motor-cam design still involves noise, but designs with linear actuators do not.

The research question this chapter focused on was:

> *Can a pump that is more compact, lighter, and more energy efficient than currently existing pumps for the portable renal replacement system be designed with lower manufacturing cost, less noise, and less possibility of clogging?*

In Chapter 6, the detail design of motor-cam design and Flexinol design will be provided, and in Chapter 7, those designs will be tested through experiments. (The ultrasonic linear motor will not be considered anymore. The lay out is the same as Flexinol design, so only one of those are being used to prove the concept. Flexinol is chosen over Ultrasonic linear motor due to the lower cost.) Chapter 8 will find the minimized pump design using the analytical model created in Chapter 5. The result will show if the concepts developed in this chapter was reasonable and the answer to the research question will finally be answered.

CHAPTER FOUR

UNDERSTANDING OF A FINGER PUMP

Chapter 3 concluded that a finger pump is the most appropriate pump for a portable renal replacement system. Although one can easily guess that the flow rate of a finger pump can be calculated by multiplying the flow amount of a finger stroke and the number of strokes per unit time, the flow amount of a finger stroke is affected by various factors such as the compliance of the tube or pump head. Thus, understanding of a finger pump is necessary before finalizing the actuator for the pump that can best provide the required finger movements.

To understand finger pumps better, this chapter first searches existing finger pump patents. Then, it explores the factors that affect the behavior of a finger pump and explains how the flow rate can be calculated. At the end, methods to increase the flow rate are suggested.

4.1 Finger Pump Patent Search

To understand finger pumps, existing patents were searched. Although Chapter 4 mostly devotes to the understanding of the flow rate of a finger pump, design characteristics of patents were noticed as well in this section. These will be considered in Chapter 5 the detail design of pump are provided.

Due to the characteristic of a finger pump that keeps the fluid inside a tube sanitary, all the patent designs searched were infusion pumps designed to replace the gravity effect of IV pole.

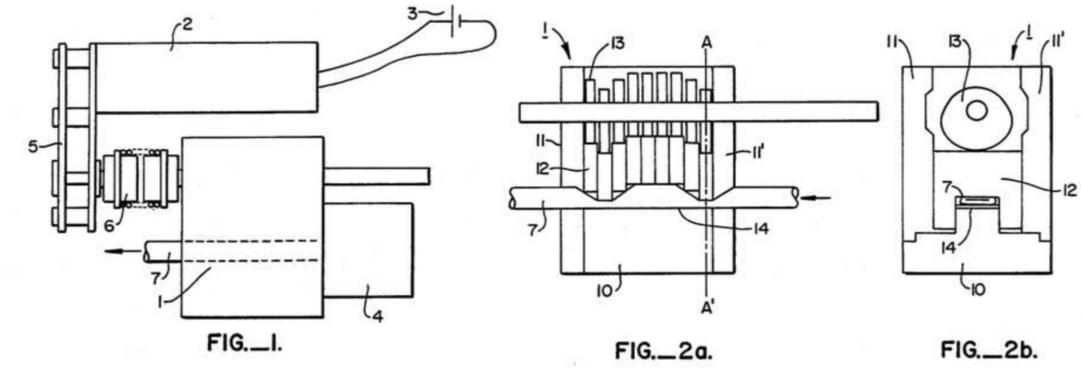


Figure 4-1. Patent no. 4952124 [11]

Figure 4-1 shows pictures of a patent that utilized two shafts, one for the motor and the other for the cams that actuates followers (fingers). The two shafts design helps reducing the size of the pump. The followers move in guides to stay in the right position. However, the guide design requires high precision manufacturing to guarantee the smooth movement of the followers without sticking. In addition, spring load is added to the followers to make sure the return to their initial positions after squeezing the tube. The spring load increases the force required to squeeze the tube which requires stronger actuation of the motor resulting in the use of a bigger motor. Thus, the guide design is not a good choice for miniaturization. The cam is designed as a disk cam that can control the compression time easily.

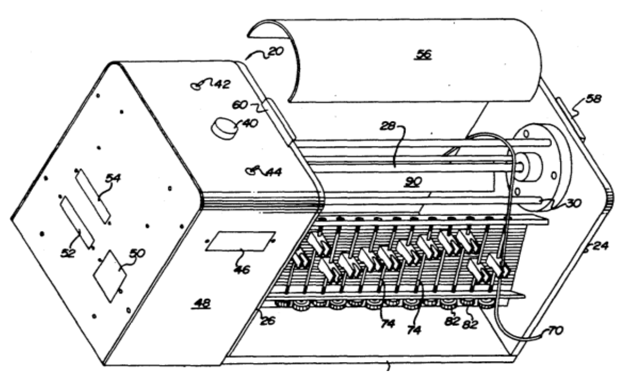


Figure 4-2. Patent no.5098261 [12]

Figure 4-2 shows a picture of a patent that controls the axial tension of each tube for the precise flow rate control. (The figure shows only the tension controlling part, and actuation part is hidden at the backside.) However, the patent design is for the accurate control in the range of few milliliters per minute of flow rate. For our design, that kind of precision is not required, so axial tension control will not be considered for a pump design. However, the fact that the axial tension affects the flow rate is noticable.

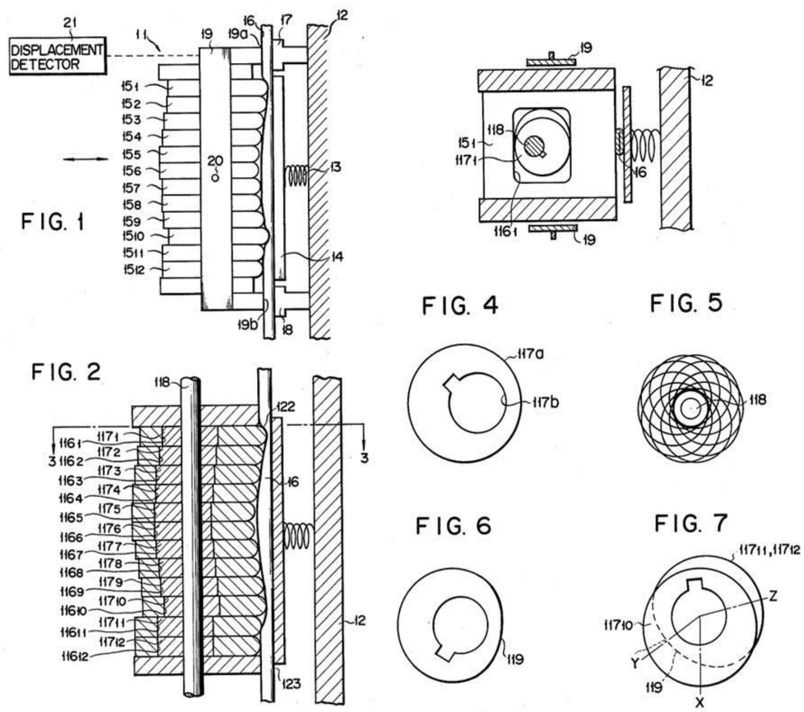


Figure 4-3. Patent no.4373525 [13]

Figure 4-3 shows a design that utilized a spring-loaded plate. Engineering tolerance could be critical for finger pumps because the fingers are the only way to stop the backflow. Small openings in the tube could lead to a much lower flow rate depending on the situation. The spring-loaded plate ensures the complete occlusion of the tube and no backflow. The patent suggests two different cam designs. One is eccentric cam design that provides stable cam follower movement, but eccentric cam will completely close the tube only for an instance. It is not recommended for a finger pump since at least one finger should be completely closing the tube at all time, and use of eccentric cam required infinite number of fingers for this. The other cam design is a common disk cam design that can easily control the duration of complete closure.

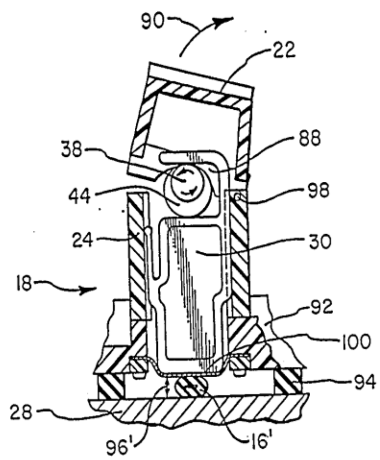


Figure 4-4. Patent no. 4725205 [14]

Figure 4-4 shows the patent that utilized a mechanism to shift the cam shaft. It is an important feature for a finger pump because positioning tube at the right spot could be very tricky. A finger pump is designed to always completely compress the tube with at least one finger. Thus, even when the tube is not inserted, one finger is in theposition of complete compression, and it provides little space for the user to insert the tube before using the pump. Such a mechanism enables an easy tube replacement. The patent also mentioned that engineering tolerances can be critical for a peristaltic pump and that the difficulty in manufacturability increases proportionately with this criticality.

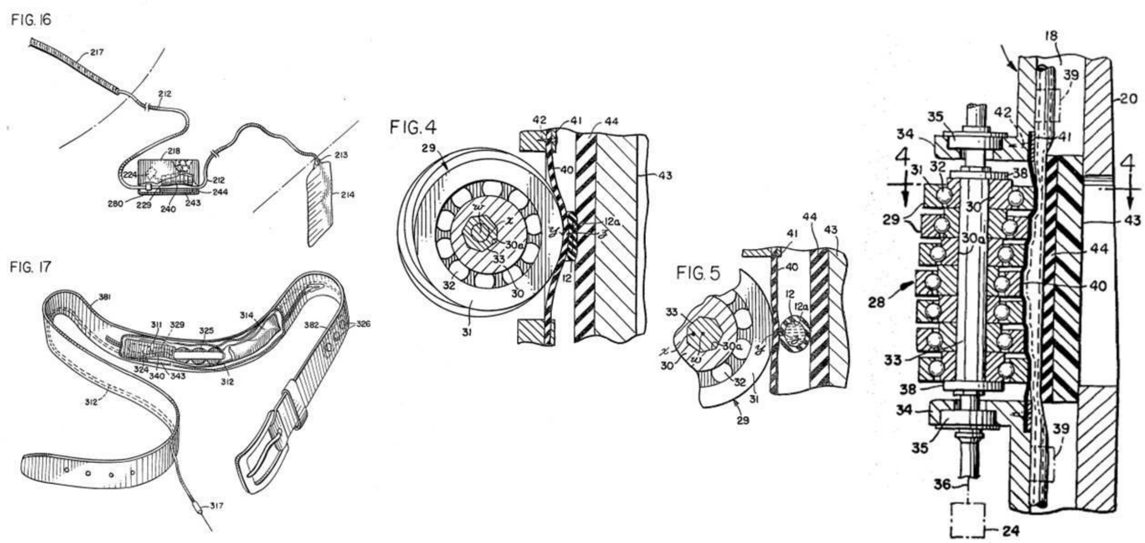


Figure 4-5. Patent no. 4482347 [15]

Figure 4-5 shows the patent that utilized a rubber plate for the same reason as patent no.4373525 explained before. This design uses a circular shaped cam follower and used ball bearing to reduce the friction loss. The circular shaped cam follower compresses the tube, and a membrane between those protects and keeps the tube at the right location stoping the tube from being pushed in one side. The patent also mentions the possibility of miniaturization for a wearable application. However, this design uses eccentric cam design that is not a good choice for a peristaltic pump as mentioned above.

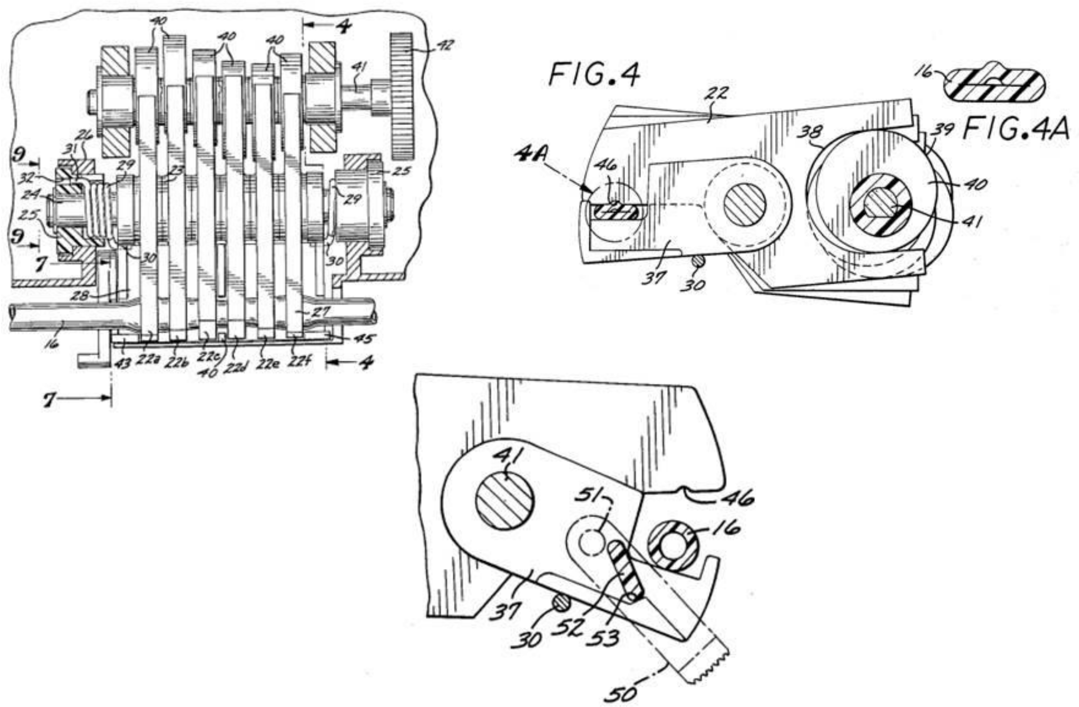


Figure 4-6. Patent no. 4561830 [16]

Figure 4-6 shows a patent that also utilized two shafts. However, this design uses lever-like designed fingers that are mounted on an axis. In addition, fingers are spaced apart from each other to reduce the contact area between each finger. Eccentric cams are used, and it has a locking mechanism to facilitate placing the tube. One particular thing to notice is the surface of the finger that meets and closes off the tube is provided with a groove in line with the tube. The patent explains that the groove is to prevent pressure buildup beyond predetermined limits. However, it will also allow backflow, so it is not recommend for the pump design for a portable renal replacement system since the backflow will reduce the efficiency of the pump.

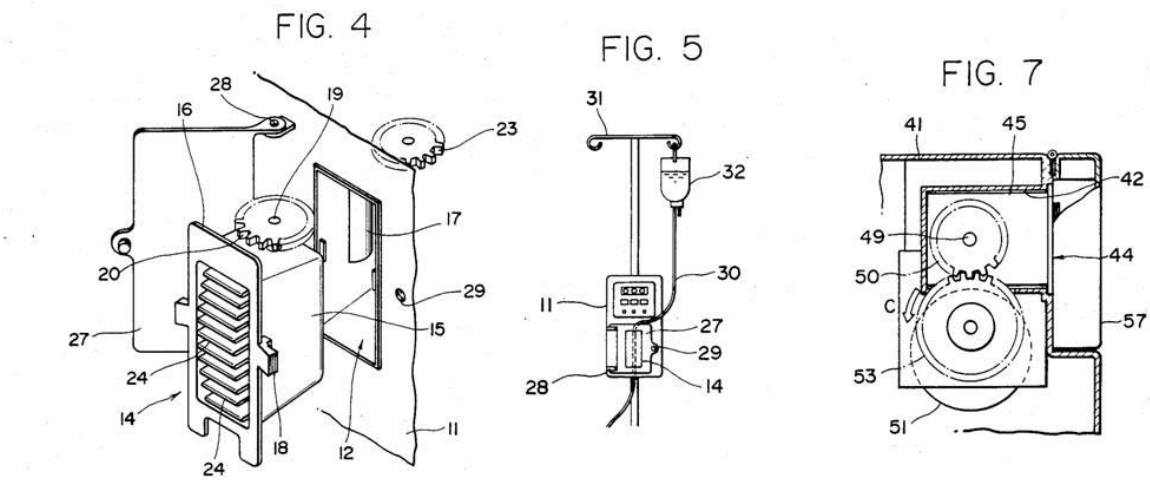


Figure 4-7. Patent no. 4653987 [17]

Figure 4-7 shows a patent design that the pump mechanism can be pulled out of the pump main body. It is a good feature for keeping the pump clean in case there is some fluid leakage or some part of the pump is broken. In addition, this pump facilitates positioning the tube by having the plate as an opening door.

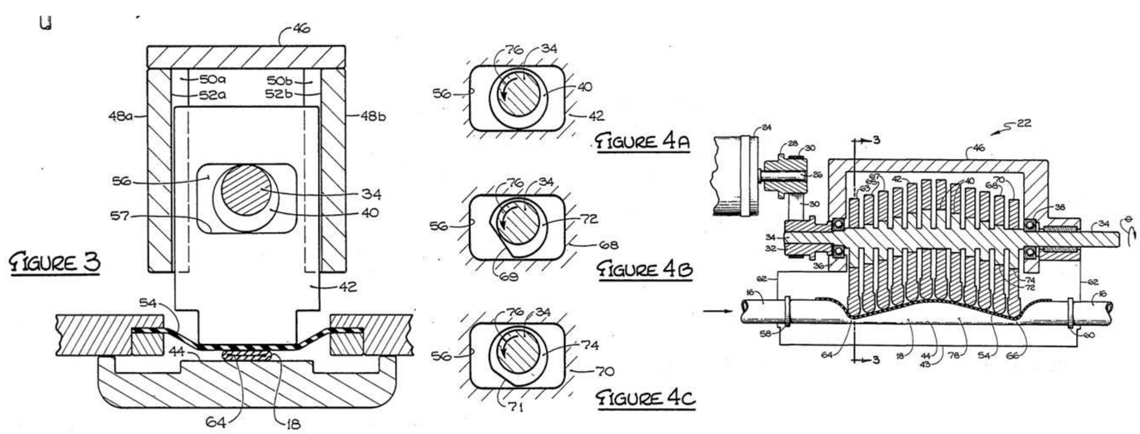


Figure 4-8. Patent no. 4909710 [18]

Figure 4-8 shows a patent that offers different cam designs that can control the compression time. Using this controllability, this design uses different compression time for the last two fingers to control the pressure increase as the compression cycle changes from the last finger to the first one. The patent also recognizes the importance of increasing the number of fingers to reduce the problems of a peristaltic pump that provides a pulsatile flow.

4.2 Governing Equations of a Finger pump

The performance of a finger pump depends on several design parameters, such as the characteristics of the tube and the actuator, which also affect the size of the pump. This section will introduce several basic equations that explain the finger pump behavior, particularly the flow rate of a finger pump.

Fingers of a finger pump sequencially compress the tube in one direction to push the fluid. The expected flow rate of a finger pump can be estimated as below.

$$\begin{aligned}&\text{flow rate (ml/min)}\\&\qquad = \text{flow amount of a finger stroke(ml)}\\&\qquad \times \text{no. of finger strokes (/min)}\end{aligned} \tag{4-1}$$

Flow amount of a finger stroke is subject to change depending on the head, friction loss, amount of back flow. Thus, it can be estimated as below.

$$\begin{aligned}&\text{flow amount of a finger stroke (ml)}\\&\qquad = \text{effective tube cross section (cm}^2) \times \text{finger width(cm)}\end{aligned} \tag{4-2}$$

Instead of using actual tube cross section area, effective tube cross section area that includes those effects are used. (More explanation will follow in the next chapter.)

No. of finger strokes per time that appears in the above equation can be calculated as below.

$$\text{no. of finger strokes (/min)} = \text{no. of fingers} \times \text{no. of cycles(/min)} \tag{4-3}$$

One cycle is from the first finger to the last finger compression. According to above two equations, the flow rate will not be affected by those two factors in RHS of (4-3) as long as the no. of finger strokes is the same assuming that the flow amount of a finger stroke is kept the same. However, it is not always true. In the next section, more in-depth explanations will specify the conditions that the flow amount of a finger stroke is kept the same.

Required flow velocity to achieve certain flow rate is calculated by dividing the target flow rate by effective tube cross section.

$$\text{required flow velocity (cm/s)} = \frac{\text{target flow rate (ml/min)}}{\text{effective tube cross section (cm}^2\text{)} \times 60} \qquad (4\text{-}4)$$

Signal delay refers to the time delay between two adjacent fingers.

$$\text{signal delay (s)} = \frac{\text{finger width (cm)}}{\text{required flow velocity (cm/s)}} \qquad (4\text{-}5)$$

If the time delay is 0.4 second, the second finger will start to compress the tube after 0.4 seconds the first finger started to compress it. The design with narrow finger width would require shorter signal delay compared to that with wide finger to achieve the same target flow rate or flow velocity because more than one finger stroke will be required during the same time duration to get the same flow rate the wide finger stroke acieved.

Above equations are the basic equations that govern the flow rate of a finger pump. At the same time, these are the design variables we need to consider.

Other than the equations explained so far, one more important factor is the force required to compress the tube. The required force is an important factor for deciding the right actuator that will provide the right finger movements to generate the required motion.

The next section will provide more in-depth explanation.

4.3 Factors that Vary the Flow Rate of a Finger pump

Although the flow rate of a finger pump is governed by the equations explained in the previous section, those equations are correct only under certain conditions. In this section, the factors that vary the flow rate of a finger pump will be introduced with suggestions to increase the flow rate.

4.3.1 Compliance of the Tube

Figure 4-9 shows the general shape of the tube as fingers compress the tube sequentially.

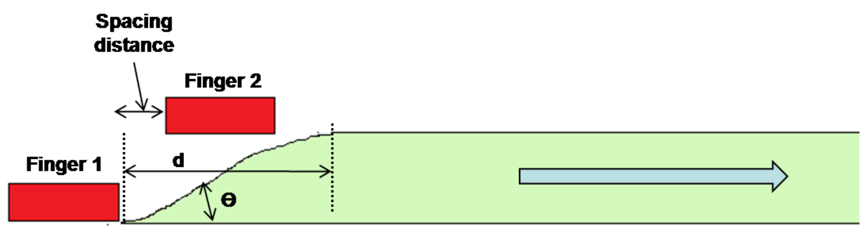


Figure 4-9. The general shape of the tube as the fingers compress the tube

As one finger compresses the tube, the tube is deformed affecting the shape of the tube more than the actual location where the finger is compressing. Thus, the compliance of the tube and the spacing distance between fingers become important factors that affect the flow rate of a finger pump.

Depending on the compliance of the tube, the angle θ is determined. If the tube is very compliant, θ is small because more neighboring areas will be affected and pulled down together. As a result, the length to recover the full tube size, d, will be long. If the tube is very stiff, θ is large making the length d short. This difference leads to the difference in the flow amount of the next finger stroke. The next finger stroke will generate different amount of flow depending on the spacing distance between two fingers. As the distance gets longer, the flow amount increases. Once the spacing distance becomes longer than the length 2d, the flow amount will not change.

Compliance of the tube also affects the time to fill the tube. If the tube is stiff, the tube can recover the original shape on its own forcing the water to fill the vacuum created. In this case, suction is possible. However, if the tube is very compliant, the pump cannot suck any fluid because the tube cannot recover its original shape on its own but requires some pressure to overcome the atmosphere pressure. In this case, the pump can only push the fluid. However, compliant tube requires less force to completely occlude the tube. Thus, when suction is not required, compliant tube can still be a good choice.

4.3.2 The First Finger Stroke Loss

Figure 4-10 shows the shape of the tube as the first finger compresses the tube.

Figure 4-10. The shape of the tube as the first finger compresses the tube

As the very first finger compresses the tube, half of the flow goes to the left and other half goes to the right because there is no valve or previous finger blocking the backflow. As a result, only half of the first finger stroke is pushed to the intended flow direction. This flow loss does not occur for other fingers due to the previous fingers compressing the tube blocking the backflow. As mentioned above, the tube compliance determines the flow amount of one finger stroke, which determines the amount of the first finger stroke loss.

4.3.3 The Last Finger Stroke Loss

Figure 4-11 shows the shape of the tube as the last finger to the first finger transition occurs. As shown, when the last finger compresses the tube, the first finger compresses the tube, and then the last one is released back while the second finger compresses the tube.

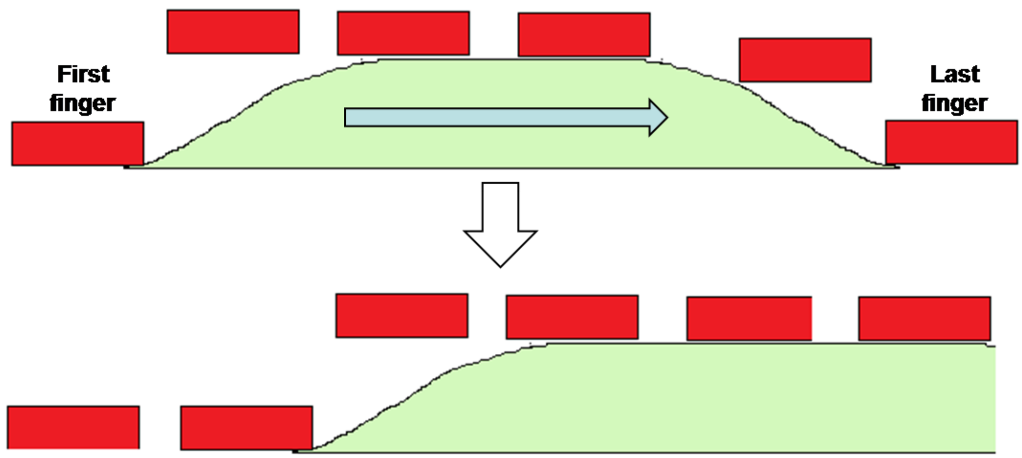


Figure 4-11. The shape of the tube as the last finger to the first finger transition occurs

As the transition occurs, some amount of backflow occurs due to the empty area created as the last finger goes up. In theory, there should be no back flow since the flow coming from the left should be filling up the gap. However, when the second finger compresses the tube, not all the flow is going forward (right) but some of the flow tends to leak through small opening under the first finger. This tendency becomes more important especially when the length of the tube compressed by fingers get smaller.

4.4 Relationship between the Compression Area and Required Force

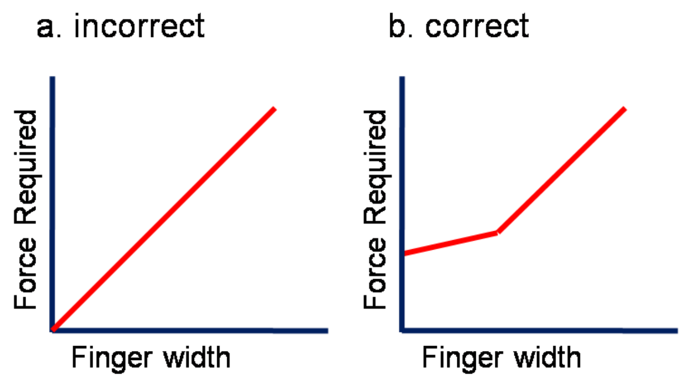


Figure 4-12. Relationship between the finger width and the force required to compress the tube

Figure 4-12 shows the relationship between the finger width and the force required to compress the tube. The left graph (a) is the wrong conception, and the right graph (b) shows the real relationship. Assuming that the cross section area is constant, the force required to compress a tube section will generally be in linear relation with the finger width. However, it is not true when the width is under certain length. For example, if 20N is required to compress 10cm of a tube, 10N will be required to compress 5cm. However, more than 1N will be required to compress 0.5cm of the tube. Thus, keeping the finger width longer than the critical length is favored for increasing energy efficiency.

4.5 Suggestions for Design

This section will introduce several suggestions to design a more efficient and smaller pump considering the factors mentioned above.

4.5.1 Compliant Tube

The use of smaller and lighter actuator can easily lead to a smaller and lighter pump design. However, smaller and lighter actuator also means less force generated. Thus, compliant tube that requires less force to compress the tube is a better choice for the design as long as suction is not required. The portable renal replacement system does not require suction because dialysate fluid can be located above the pump to provide the pressure head to fill the tube. For the blood circuit, the blood pressure will provide enough pressure to fill the tube. As a result, this pump will be using a very compliant tube, the penrose drain tube.

4.5.2 Spacing between Fingers

When the total length of the tube that is compressed by fingers is 9cm, the flow rate can be the same when either nine 1cm width fingers are compressing with no spacing in between or five of the same fingers are compressing with 1cm spacing in between. However, the latter design requires compression times twice as long as the former design for each finger to ensure that the tube is completely occluded all the time. Thus, the required force for an actuator is the same. As a result, the spacing between fingers does not affect the size or weight of the pump too much. However, the spacing can still be important because it will be able to reduce the friction between fingers. If the fingers have to be assembled right next to each other, there will be friction loss. The spacing can reduce the friction loss by providing additional component (spacer) between fingers. Less friction loss will lead to a more efficient design.

4.5.3 Length of the Tube

As mentioned above, both the first and the last finger strokes pump less fluid than other fingers. Thus, reducing the number of cycles will minimize the loss. As a result, a longer tube to be compressed by fingers will yield higher flow rate. However, it increases the size of the pump at the same time. For the design of the pump, these two factors will be considered to find the best length of the tube for the pump.

4.5.4 Number of Fingers

As metioned in the previous section, the number of fingers should not affect the flow rate of the pump as long as the flow velocity is the same. In other words, having four 2cm fingers with 1 second signal delay is theoretically the same as having eight 1cm fingers with 0.5 sec signal delay because both have the same fluid velocity, 2cm/sec.

Three fingers are required to keep the reasonable finger pump operation. (Three fingers can make sure the complete closure of at least one finger all the time while keeping constant fluid transport in one direction.) However, as mentioned above, some time is required to fill the tube. Not just longer tube, but increasing the number of fingers can also help providing enough time for it because only three fingers will be working at any instant. For example, consider pump designs that run six cycles per minute. And consider the instant the last finger completely compresses the tube and starts the next cycle with the first finger starting compression. If a pump has four 2 cm width fingers, the first 6 cm of the tube were open for less than 7.5 seconds depending on the duration time of each finger. If a pump has eight 1cm width fingers, the first 7 cm of the tube were open for less than 8.75 seconds. This example shows how having more fingers can increase the flow rate by increasing the time to fill up the tube. However, increasing the number of components could involve increase in manufacturing and maintenance cost, so the right number of fingers will have to be chosen.

4.5.5 Signal Delay and Duration Time

Providing enough time to fill up the tube is very important when a compliant tube is used. The length of the tube and the number of fingers are important because of this. One more way that could be helpful is controlling the signal delay and the duration time especially for the first and the last fingers. Signal delay is the time delay between two adjacent fingers, and the duration time is the time a finger is completely occluding the tube allowing little back flow. Rather than using the same signal delay and duration time for all the fingers, applying different values for the first and the last fingers could increase the flow rate by providing more time to fill up the tube and reducing the back flow.

CHAPTER FIVE

CREATION OF AN ANALYTICAL PUMP MODEL

The finger pump study in the Chapter 4 provided the basic understanding of a finger pump. This chapter will explore all the design parameters of a finger pump and develop an analytical pump model in Simulink. The pump model provides the expected pump size and performance for a given configuration and actuator characteristics. Then, the effect of input variable changes on the output variables will be investigated to provide a guideline for miniaturizing pump designs in the later chapters.

5.1 Motivation and Requirements Lists

The performance of a finger pump depends on several design parameters, such as the characteristics of the tube and the actuator, which also affect the size of the pump. These design parameters have simple relationship between each other. However, many parameters are inter-related with each other making it hard to predict how the pump will perform for a given configuration and actuator characteristics. Thus, an understanding of the relationship of these parameters is crucial for the systematic minimization of the pump and finalization of the actuator for the pump.

The requirements list will delineate the demands and wishes from the pump model that shall be created. Table 5-1 provides the requirements for the pump model.

Table 5-1. Requirements for the pump model

D/W		Requirements Responsible: Jane Kang
D	1.	**Include every quantifiable design parameters that affects the design**
D	2.	**Quantify the design output values** : The design output values will include the expected power consumption and the total size of the pump.
D	3.	**Adaptable for different flow circuit** : The model should be adaptable for different pump head, friction loss, etc.
D	4.	**Easy to change variable values** : To find the best design for the system, changing the values should be easy enough to try many different values.
W	5.	**Adjustable for different requirements** : The model should be useful for designing different pumps such as the target flow rate of 100ml or 1000 ml per minute.

5.2 Exploring the Primary Parameters' Relationships

This section will explore relationships of primary parameters that are included in the pump model and affect the pump design. Although some additional parameters might be required depending on the characteristic of actuators, the primary parameters will not change because these are the common ones for finger pumps. First, the relationships will be explained, and all the parameters will be presented for the pump model.

5.2.1 Experimental Setting and Tube Characteristics

The cross section of the tube is defined as the section normal to the flow direction. The cross section of tube is circular, but when the tube is inserted to the pump, its cross section becomes a long oval that fits a right trapezoid space between fingers and plate. (Figure 3-10) To make the equation simpler, the tube is abstracted as a long oval with long side of tube width and other short side of squeeze distance.

$$\begin{aligned}&\text{effective tube cross section } (\text{cm}^2)\\&\qquad= \text{oval constant} \times \text{tube width(cm)} \times \text{squeeze distance(cm)}\end{aligned} \tag{5-1}$$

Peristaltic pumps are subject to head change, thus the model should be able to consider those different cases. The oval constant can be used as a lumped constant that takes account of pump head change, back flow loss, and friction loss. Friction loss would include the effect of the length of the flow circuit and inner diameter of the tube. Default value for the oval constant is π/4=0.785 since this number is the ratio of a oval area that fits a rectangle to the rectangle. The tube width change that happens during the pump operation due to the squeezed tube is ignored since it will stay almost the same (because the squeeze distance is much shorter than the tube width). To determine the oval constant, an experiment can be conducted to measure the flow rate, and the oval constant can be used to match the experimental data with the model prediction. For example, if the pump is operating with high disposal side head, the flow rate will decrease. In that case, the oval constant will be smaller than the default value.

The compliance of the tube is the factor that provides the requirement of an actuator. Quantifying the compliance of the tube can be done by specifying the force required to squeeze the tube.

5.2.2 Design Requirements

The model is expected to be useful in designing different pumps with different target flow rates. Thus, the target flow rate of the pump is one of the design parameters of the model. However, this flow rate is not the only flow rate that the pump can generate. For the design process, the maximum value can be used, and the flow rate can be varied by varying signal delay time or cycle/min, or size of the tube or tube width. The relationship between the tube characteristics and the target flow rate is already explained in Chapter 4.

$$\text{required flow velocity (cm/s)} = \frac{\text{target flow rate (ml/min)}}{\text{effective tube cross section (cm}^2\text{)} \times 60} \qquad (4\text{-}4)$$

Fluid power refers to the theoretically calculated power required to transport the fluid at a given flow rate with a flow pressure.

$$\text{fluid power (W)} = \text{Flow Rate} \times \text{Flow Pressure} \qquad (5\text{-}2)$$

If we assume the blood pressure of 100mmHg, we can simplify the equations by relating to the target flow rate as below. [5]

$$\text{fluid power (W)} = 0.0222 \times \text{target flow rate (ml/min)} \qquad (5\text{-}3)$$

Finger width is the length of the tube that can be occluded each time a finger compresses. It affects the size of the pump with the number of fingers, and the relationship will be explained in the later section.

The flow amount of a finger stroke, no. of finger strokes, and signal delay were explained in Chapter 4 as well.

$$\begin{aligned}&\text{flow amount of a finger stroke (ml)}\\ &\qquad = \text{effective tube cross section (cm}^2\text{)} \times \text{finger width(cm)}\end{aligned} \qquad (4\text{-}2)$$

$$\text{no. of finger strokes (/min)} = \text{no. of fingers} \times \text{no. of cycles(/min)} \qquad (4\text{-}3)$$

$$\text{signal delay (s)} = \frac{\text{finger width (cm)}}{\text{required flow velocity (cm/s)}} \qquad (4\text{-}5)$$

Operating time is determined depending on the treatment hours. It is important information that will be used for finding the right battery since the battery should be able to run the pump for one whole treatment. The relationship will be explained in the next section.

5.2.3 Actuator Characteristics

An actuator provides the force required to squeeze the tube. Thus, the force an actuator can generate is important information for choosing the right actuator. The required force generation depends on the tube characteristics and design requirements.

Eqn (5-4) shows how to calculate the required force generation per finger. This information is useful for the design with linear actuators because each actuator will be actuating each finger. However, the total required force is more important for the motor cam design because one motor will be actuating all the fingers. In this case, Eqn (5-5) wil be more useful for finding the right actuator for the design.

$$\begin{aligned} &\text{required force generation per finger (N)} \\ &\quad = \text{force required to squeeze (N/cm)} \times \text{finger width (cm)} \end{aligned} \qquad (5\text{-}4)$$

Eqn (5-5) shows how the total required force can be calculated. Because it is for a finger pump, not all but only about three fingers are squeezing the tube at an instant. Thus, not the number of fingers but the number of working fingers at an instant is multiplied in the equation.

$$\begin{aligned} &\text{total required force generation (N)} \\ &\quad = \text{required force generation per finger (N)} \\ &\quad \times \text{no. of working fingers at an instant} \end{aligned} \qquad (5\text{-}5)$$

Each actuator has different resistance and operates at different voltage to generate the force to squeeze the tube. This information will totally depend on the data sheet of the actuator and will be used for providing power consumption information. The relationship between the generated force and power will vary depending on the actuators. However, Eqn (5-6) will always hold.

$$\text{Voltage (V)} = \text{current (A)} \times \text{resistance (Ohm)} \tag{5-6}$$

5.2.4 Power Consumption

Power requirement is an important factor that decides the weight of the total system. Thus, the power consumption information is one of the primary parameters that helps the evaluation process of different designs.

Ampere hour refers to the required battery capacity. This information is required for the search of the right battery with voltage information. If the pump needs to operate longer, more capacity is required.

$$\text{Ampere hour (Ah)} = \text{current (A)} \times \text{operating time (hr)} \tag{5-7}$$

Watt hour is also required information to evaluate the weight of the required battery. In section 3.2.5.4, it was shown that the weight of the battery linearly increases with the required watt hour.

$$\text{Watt hour (Wh)} = \text{Ampere hour (Ah)} \times \text{V applied (V)} \tag{5-8}$$

Brake power refers to the power required to operate the pump. It is calculated by multiplying the required battery voltage and current.

$$\text{brake power (W)} = \text{voltage (V)} \times \text{current(A)} \tag{5-9}$$

Efficiency refers to the pump efficiency. It is calculated by dividing the theoretical output power, fluid power, by practically required power, brake power, to pump the fluid.

$$\text{efficiency} = \frac{\text{fluid power (W)}}{\text{brake power (W)}} \tag{5-10}$$

5.2.5 Pump Size Infromation

The size of the pump is one of the most important requirements to the evaluation of a design. Figure 5-1 shows the layout of a pump design with marks that explain how the pump depth, width, and height are defined.

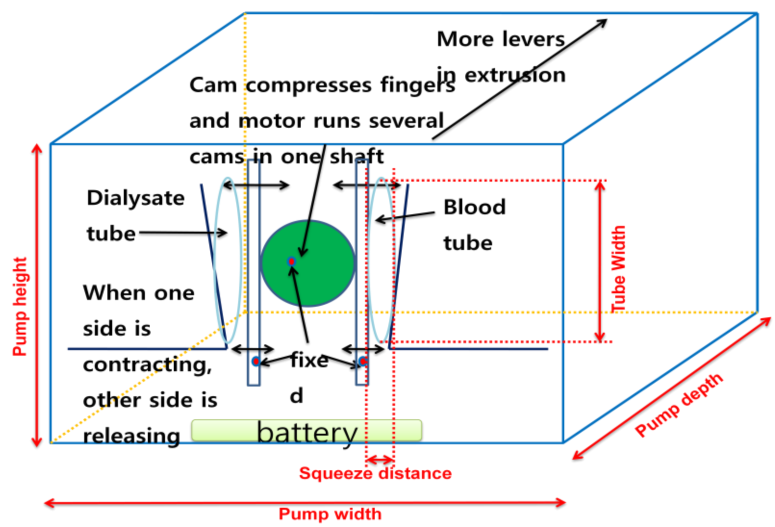


Figure 5-1. Layout of a finger pump design (with pump length mark)

Since the pump depth is defined as the length of the pump in the flow direction, the relationship can be expressed as below.

$$\text{pump depth (cm)} = \text{ no. of fingers} \times \text{finger width (cm)} + \alpha \tag{5-11}$$

The width of the pump is defined as the same direction as the sqeeze distance (shorter side of the tube). The pump is left-right symmetric to pump two different fluids, so the relationship can be expressed as (5-12).

$$\text{pump width (cm)} = 2 \times [\text{squeeze distance (cm)} + \beta] \tag{5-12}$$

The height of the pump is defined as the same direction as the tube width (longer length of the tube), so the pump height can be estimated by adding tube width and additional length γ.

$$\text{pump height (cm)} = \text{tube width (cm)} + \gamma \tag{5-13}$$

α, β, and γ are the additional length due to the frame of the pump. This information will have exact values after designing an actual pump with a certain setting. Then, the values can adjust to the design change.

The body volume of the pump can be calculated by multiplying three lengths as below.

$$\text{pump body volume (cm}^3) = \text{depth (cm)} \times \text{width (cm)} \times \text{height (cm)} \tag{5-14}$$

Section 3.2.5.4 introduced the below equation that relates the power consumption to the size of required battery.

$$\text{battery volume (cm}^3) = \text{battery size constant (cm}^3/\text{Wh)} \times \text{Watt hour (Wh)} \tag{3-2}$$

Thus, the total estimated pump size can be the sum of those two volumes.

$$\text{total volume (cm}^3) = \text{pump body volume (cm}^3) + \text{battery volume (cm}^3) \tag{5-15}$$

By converting the power consumption information to the battery volume, the ultimate goal of the pump design reduced to minimizing the total volume, not mimizing the body volume while increasing efficiency to reduce the size of the battery.

5.2.6 Primary Parameters of the Analytical Pump Model

Table 5-2 shows the primary parameters introduced in the previous sections. Primary parameters are the ones commonly used for any actuator. Each parameter is designated as either an input, intermediate calculation, or output. Input paraters are type of tube, requirements, and actuator characteristics. Output parameters provide information of the pump design specifications. Intermediate parameters are used to get the output values from input values.

Table 5-2. Primary parameters of the analytical pump model

	Parameters	type		
		Input	Intermediate parameters	Output
Experimental Setting	Oval constant	◎		
Tube Character	Effective tube cross section		◎	
	tube width (cm)	◎		
	squeeze distance (cm)	◎		
	Force required to squeeze (N/cm)	◎		
Design Requirements	target flow rate (ml/min)	◎		
	fluid power (W)			◎
	Operating time (hr)	◎		
	Required flow velocity (cm/s)		◎	
	finger width (cm)			◎
	No. of fingers	◎		
	No. of fingers working at an instant	◎		
	Flow amount of a finger stroke (ml)		◎	
	No. of cycles per minute (/min)			◎
	No. of finger strokes (/min)		◎	
	signal delay (s)			◎
Actuator Character	Required force generation per finger (N)			◎
	Total required force generation (N)			◎
	Voltage (V)	◎		
	Resistance (Ohm)	◎		
	Current (A)			◎
Power Consumption	Ampere hour (Ah)			◎
	Watt hour (Wh)			◎
	Battery size constant (cm^3/Wh)		◎	
	brake power (W)			◎
	Efficiency			◎

Table 5-2 continued

Pump Size Information	pump width (cm)	◎
	pump height (cm)	◎
	pump depth (cm)	◎
	pump body volume (cm^3)	◎
	Battery volume (cm^3)	◎
	Total volume (cm^3)	◎

5.3 Exploring Design Parameters for Different Actuators

5.3.1 Design Parameters for Motor-Cam Driven Pump Design

No additional parameters are required for motor cam design because the actuator characteristics of the motor follow the simple relationship that is embedded in the base model. Thus, the pump model for the motor will be the same as the base model.

5.3.2 Design Parmeters for Flexinol Design

Contraction distance refers to the Flexinol contraction distance. Because the lever mechanism is used, the Flexinol contraction distance is shorter than the squeeze distance when the lever ratio is higher than 1. For example, if required squeeze distance is 0.8 cm and lever ratio is 8, only 0.1mm of Flexinol contraction distance is needed.

$$\text{contraction distance (cm)} = \frac{\text{squeeze distance (cm)}}{\text{lever ratio}} \tag{5-16}$$

Flexinol length refers to the required Flexinol length to get the required contraction distance. The contraction ratio of the Flexinol can go up to 5% without damaging it (data sheet). If a designer choose to contract 4% of Flexinol to be safe, 10 cm Flexinol length would be needed to get 4mm of contraction distance.

$$\text{flexinol length (cm)} = \frac{\text{contraction distance (cm)}}{\text{contraction ratio}} \tag{5-17}$$

The resistance refers to resistance of each Flexinol. It can be calculated by multiplying the Flexinol length and unit resistance, which is given as a characteristic of the Flexinol.

$$\text{resistance} = \text{unit resistance (ohm/cm)} \times \text{flexinol length (cm)} \qquad (5\text{-}18)$$

Flexinol efficiency in Eqn (5-19) refers to the kinetic energy it can generate divided by the electrical energy required to generate it. From this equation, the current that is required to contract the Flexinol during the given contraction time can be calculated.

$$\text{current}^2 \times \text{resistance} \times \text{flexinol efficiency (W)} = \frac{\text{Force } F \text{ generated per flexinol (N)} \times \text{contraction distance (m)}}{\text{contraction time (s)}} \qquad (5\text{-}19)$$

V required refers to the voltage that should be provided to get required current that will contract the Flexinol in given contraction time. It is calculated by multiplying current and resistance of each Flexinol.

$$\text{V required (V)} = \text{current} \times \text{resistance} \qquad (5\text{-}20)$$

No. of Flexinol/finger refers to number of Flexinols that should be connected to each finger. It increases as the finger width or lever ratio increase because more force is required. If a Flexinol that can generate more force is used, no. of Flexinol/finger will decrease.

$$\text{no. of flexinol/finger} = \frac{\text{Force } F \text{ required to squeeze (N/cm)} \times \text{finger width (cm)} \times \text{lever ratio}}{\text{F generated/flexinol (N)}} \qquad (5\text{-}21)$$

Ampere hour refers to the required battery capacity. If it needs to operate longer, more capacity is needed.

$$\text{Ampere hour (Ah)} = \frac{\text{no. of flexinol/finger} \times \text{current (A)} \times \text{total contraction time (s)} \times \text{operating time (hr)}}{\text{signal delay (s)}} \qquad (5\text{-}22)$$

Number of fingers required is calculated by dividing the sum of cooling time and squeezing time by signal delay. The sum of cooling time and squeezing time is the time it takes for a finger to operate one cycle of compressing and releasing. Because signal delay is the operating time delay between two adjacent fingers, the equation is giving the minimum number of fingers that is required to achieve required fluid velocity without damaging the Flexinol by heating before it cools down.

$$\text{no. of fingers required} = \frac{\text{cooling time (s)} + \text{ squeezing time (s)}}{\text{signal delay (s)}} \qquad (5\text{-}23)$$

Table 5-3. Additional parameters of the analytical pump model (Flexinol)

Parameters	type			
	Additional Input	Intermediate parameters	Output for the primary input	Additional Output
Lever ratio	◎			
Contracton ratio	◎			
Contraction distance (cm)		◎		
Unit resistance (Ohm/cm)	◎			
Resistance per Flexinol (Ohm)			◎	
Flexinol length (cm)				◎
Flexinol Efficiency	◎			
Squeezing time	◎			
Cooling time	◎			
No. of Flexinol per finger				◎
F generated per Flexinol (N)	◎			
Required F generation per finger				◎
No. of cycles per minute (/min)			◎	
Voltage required (V)			◎	
Mosfet Current				◎

5.4 Developing the Analytical Pump Model in Simulink

This section will create the pump model in Simulink based on the parameter relationships explained in the previous section.

5.4.1 Create the Base Finger Pump Model (Motor-Cam Driven Pump)

Based on the primary parameters' relationships, the base finger pump model is created in Simulink as shown in Figure 5-2, and the bigger version is attached in Appendix.

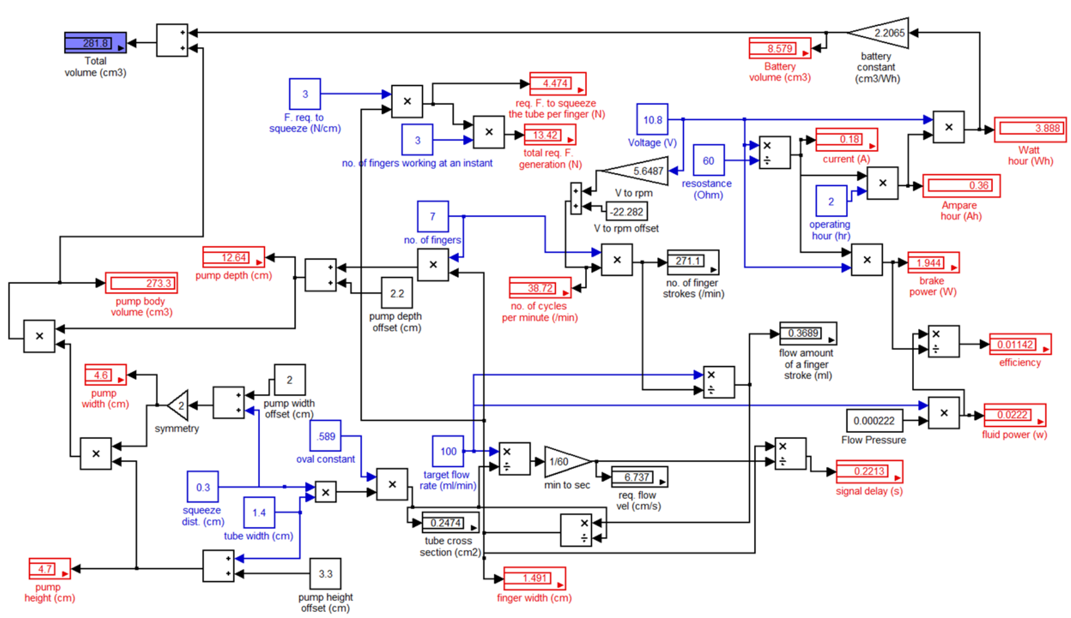


Figure 5-2. Base finger pump model in original form

Inputs are shown in blue, outputs are in red, and intermediate parameters are in black. The purple filled box on the left top is the total volume of the pump, which is the ultimate output of the pump model. Since these parameters are all interrelated, it is not easy to read. To make the model easier to read and use, the base model is organized using subsystem hiding all the intermediate parameters and calculation blocks as shown in Figure 5-3.

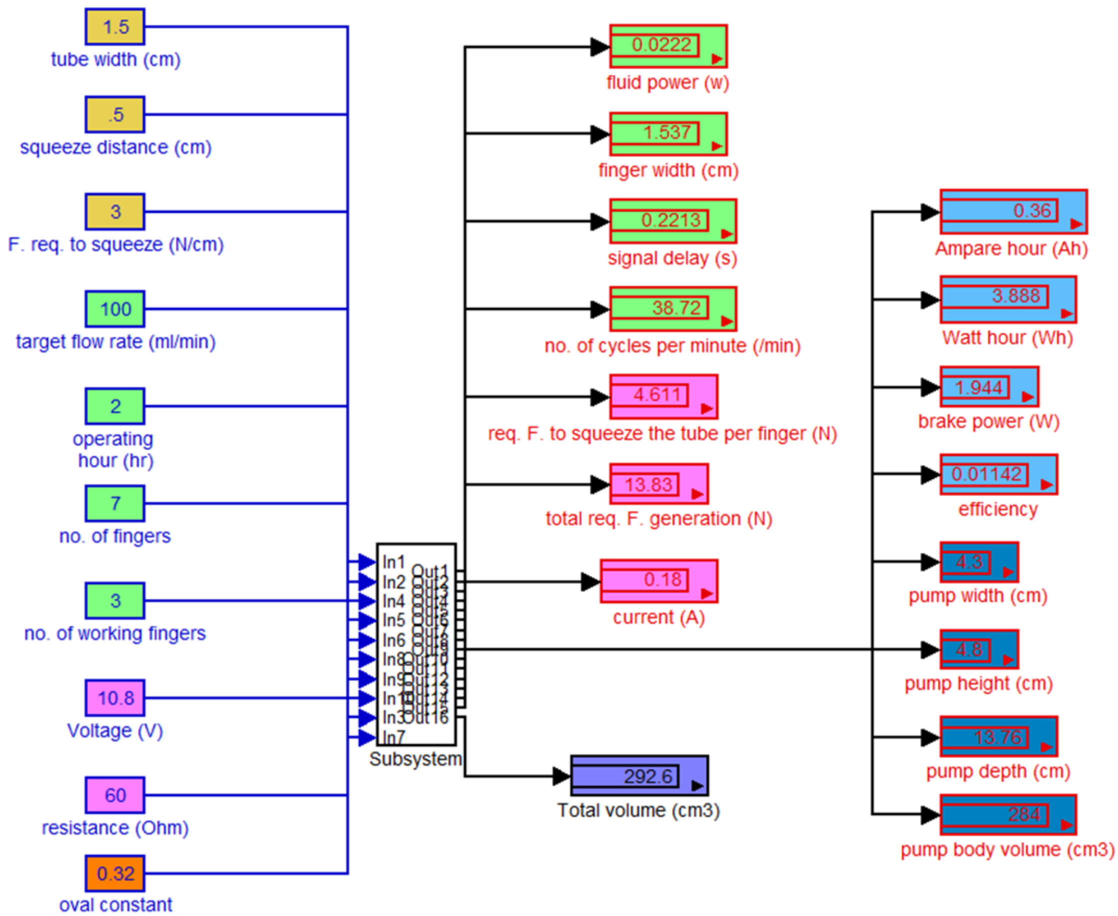


Figure 5-3. Base finger pump model with subsystem

Figure 5-3 shows inputs on the left and outputs on the right so that a user can enter values in the left and look at the results on the right. The parameters in the same categories have the same background color, tube characteristics are in yellow (top three on the left), design requirements are in green (next four on the left and top four on the right first column), actuator characteristics are in pink (next two on the left and next three on the right first column), oval constant in orange (bottom on the left) power consumption are in light blue (top four on the right second column), pump body size information in blue (bottom four on the right second column), and total volume in purple (bottom on the right first column).

From the input values given on the left, the model will calculate and provide output values on the right. In this case, when the tube width is 1.5cm and squeeze distance is 0.5cm with the force required to squeeze 3N/cm, the total volume of the pump will be 292.6cm3 when 10.8V is applied to operate the motor to generate 100ml/min of flow rate with 0.32 oval constant.

5.4.2 Adjust the Base Model for a Different Actuator (Flexinol)

To create the Flexinol driven pump model, the base model is slightly modified with additional blocks related to the actuator charactors.

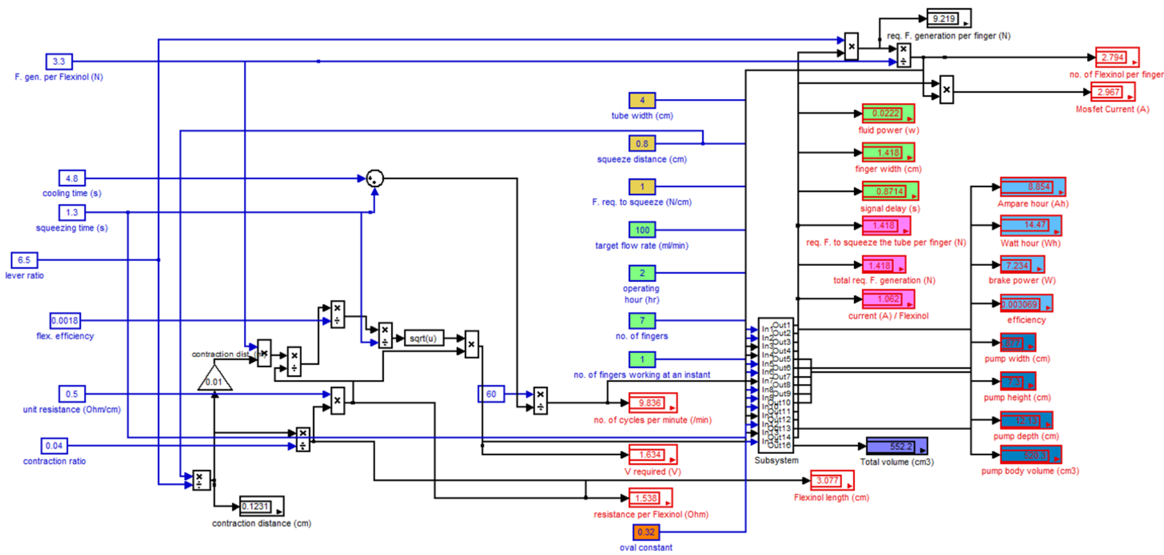


Figure 5-4. Adjusted finger pump model in original form (Flexinol)

Figure 5-4 shows adjusted design model which implemented additional design parameters introduced in 5.3.2. (The bigger version will be available in the Appendix.) The additional blocks are shown on the left and connected with the parameters already built in Figure 5-3. Several of inputs in Figure 5-3 are replaced with additional block's outputs. As before, to make it easier to read and use, subsystem is used as shown in Figure 5-5, and the bigger version will be provided in Appendix.

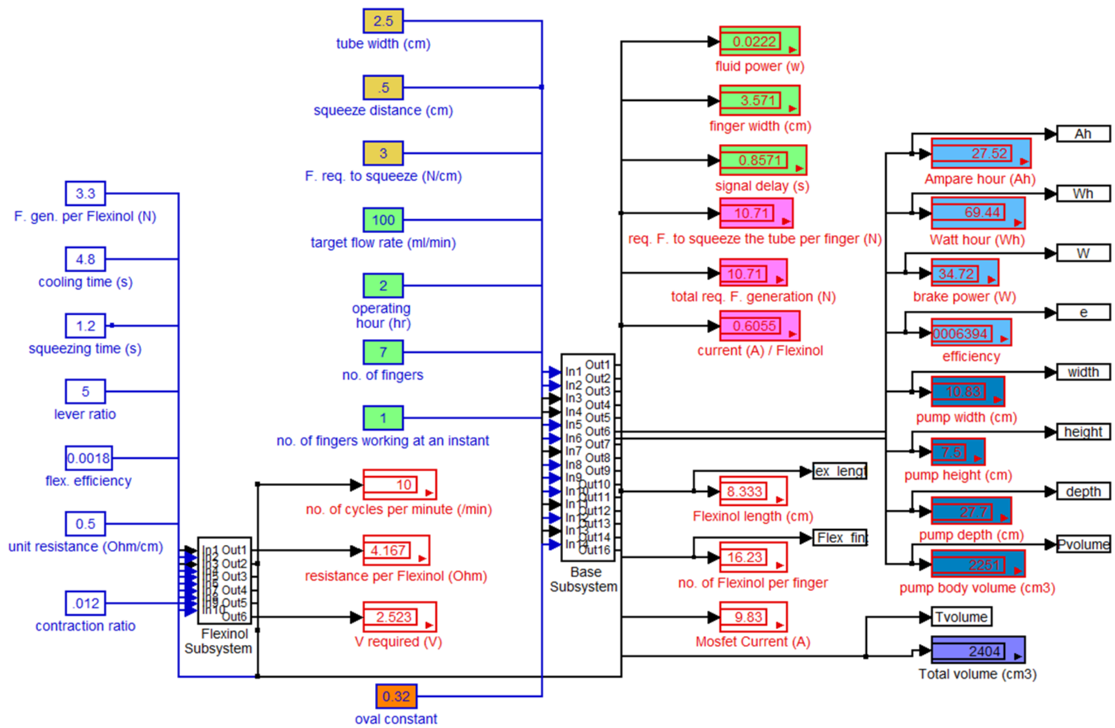


Figure 5-5. Adjusted finger pump model with subsystem (Flexinol)

Blue boxes with no fill color on the first column are the additional input parameters that include Flexinol characteristics and lever ratio. From the additional input characteristics, which are found from Flexinol data sheet, boxes with no fill color (red boxes) located in the middle two columns were calculated. These boxes on the second column are input values for the base subsystem. These boxes on the third column are the additional outputs due to the use of a different actuator.

5.5 Assumptions and Limits of the Analytical Pump model

The pump model created above started with several assumptions, and it can predict the pump performance under certain conditions. This section will explain those assumptions.

- The model assumes that the flow amount of a finger stroke is always the same. Chapter 4 explained that the amounts of the first and last finger stroke are different from other middle fingers. However, this model did not take account of the difference for simplicity.
- Compression sequence could be changed by varying the cam design or signal delay for each finger. However, the model assumes that no such difference exists. As long as the average value is used, the result will be close enough.
- The motor-cam driven pump model assumes that the motor fits the space determined from the offset for width, length, and depth. Thus, the size change according to the change in the force requirement for an actuator is not embedded. It could be done by analyzing the trend of the motor size change due to the change in the force requirement. It was done for the Flexinol driven pump model since the information is given.
- For the motor-cam driven pump model, the relationship between the force generation and power consumption is not embedded due to lack of information. It could be added once the equation is provided, but in this model, an average resistance value will be used to predict the power consumption. It is embedded in the Flexinol design.
- As explained in Chapter 4, the force required to squeeze a tube does not increase linearly when the compression area increases, especially when the compression area gets smaller. However, the model assumes that those are in linear relation for simplicity. Thus, the designer will have to consider the factor and keep the width of the finger longer than 0.5cm.

5.6 Understanding How the Input Variables Affect the Output Variables

This section will investigate the effects of input variable change on the output variables. To understand the effects, input values are set to a constant that is believed to be a close value to an optimized design. Then, one input variable is changed each time, and the output values and the input values are graphed accordingly. It should be noted that the trend of the output variables is more important than the value itself because the values will change if any of the other input values change, but the trend will stay the same. In addition, the trend might be different in different input value ranges. However, a designer's interest will only be in the region that could be implemented in the actual pump. Thus, investigating small range of input variable values will be good enough as long as the range includes the minimum and the maximum possible values.

5.6.1 Motor-Cam Driven Finger Pump Model

Output changes for the motor-cam driven finger pump model are observed first. Figure 5-6 shows the output changes according to the tube width change.

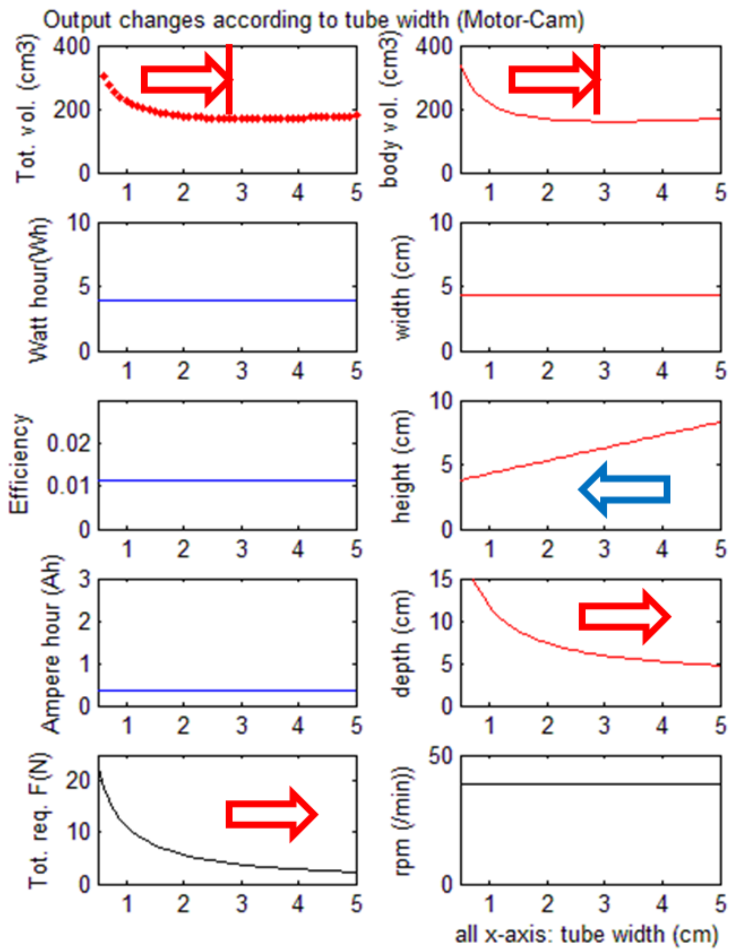


Figure 5-6. Output changes according to tube width (Motor-Cam)

The first row, first column that is colored in red with dots is the total pump volume change. Since the total volume of the pump is addition of pump body volume and battery volume, and the battery volume is very small compared to the pump body volume, usually total volume follows almost the same pattern as pump body volume. In this case, since Watt hour does not change according to the tube width change, total volume has exactly the same trend as pump body volume with a little addition in the volume for the battery volume.

The graphs colored in blue are graph of the output values related to power consumption and efficiency of the pump. As explained above, efficiency is calculated by dividing fluid power by brake power. Watt hour and ampere hour are the information required to know when looking for the correct battery. As mentioned before, the factor that decides the size of a battery is Wh, so reducing Wh is more important than reducing Ah.

The top four rows of the second column colored in red are graph of the output values related to the body size of the pump. Reducing the volume is the goal, but the width, height, and depth are plotted as well to facilitate the understanding of the finger pump design.

The bottom rows colored in black are graph related to the requirements of an actuator. Less total required force generation is better because it affects the size of an actuator (in this case, a motor).

Above figure shows that the tube width does not affect the efficiency of the pump. However, it affects the volume of the pump and the total required force generation for an actuator. The arrows in red indicate that the tube width is desired to be larger for a favorable output value and the blue arrow is the opposite. In this case, shorter tube width will reduce the height of the pump, but longer tube width will reduce depth and total required force generation. The volume of thetotal pump will decrease as the tube width increases up to a certain point, and increase again when the tube width is too long. This is due to the opposite trend of the height and depth of the pump. The effect of decreasing depth is dominant when the tube width is short, but the inceasing height becomes dominant when the tube width becomes too long. Thus, the tube width should be increased up to a certain point.

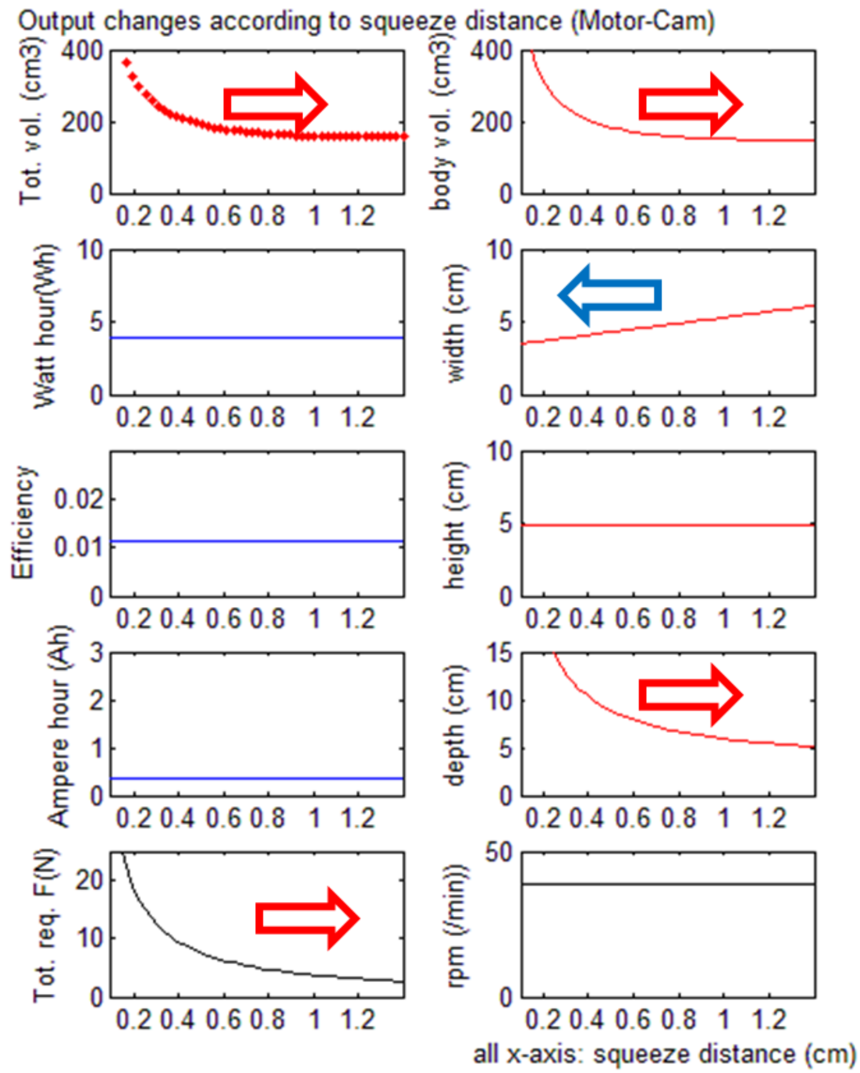


Figure 5-7. Output changes according to squeeze distance (Motor-Cam)

Figure 5-7 shows the output changes according to the squeezing distance change. It suggests increasing the squeezing distance since it decreases both volume and required force generation while not affecting the efficiency of the pump or power consumption.

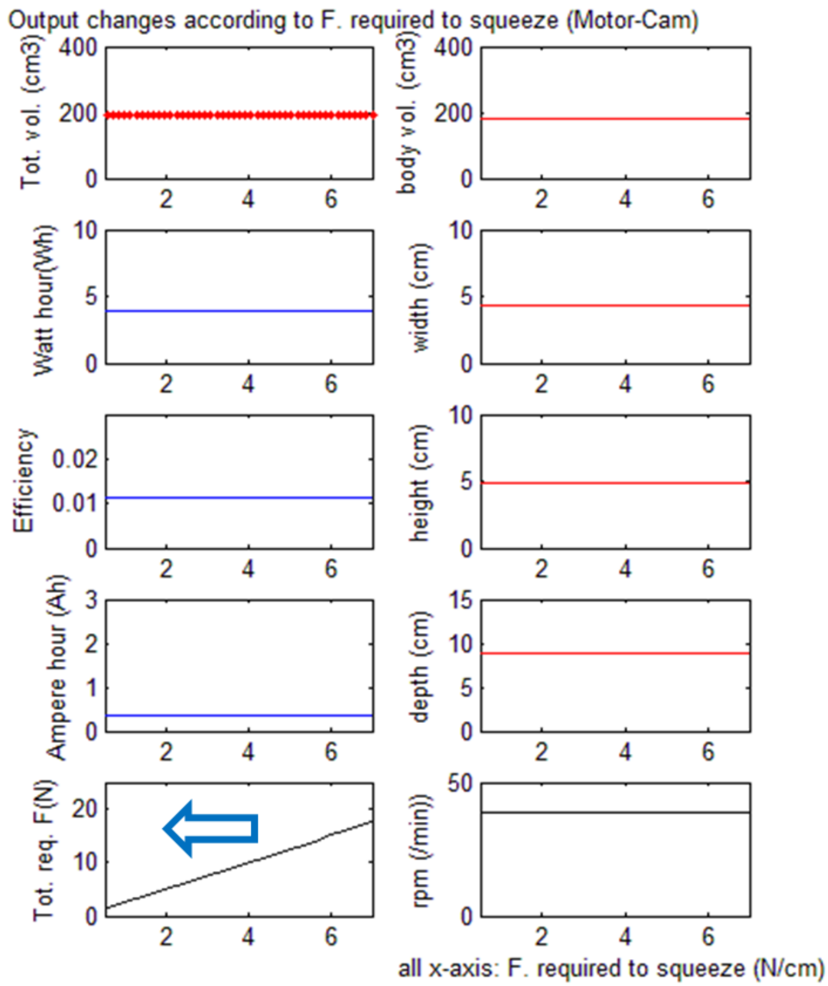


Figure 5-8. Output changes according to force required to squeeze (Motor-Cam)

Figure 5-8 shows the output changes according force required to squeeze, which is the measure of the tube compliance. It shows that only the total required force generation is affected. However, as mentioned in the previous section, the model did not take account of the relationship between the force generation and power requirements. For example, consider a motor that rotates a cam shaft to squeeze a tube. If a stiffer tube is used, although the same voltage is applied, the current will increase, thus increasing power consumption and decreasing efficiency. The relationship could be implemented in the model once the equation is given. Even with the fact that the relationship is not embedded in the model, it does not affect the designer's decision too much since the arrow direction will be the same for the total required force generation and efficiency. The result leads to the same conclusion from Chapter 4 and suggests using a compliant tube to reduce the force required to squeeze.

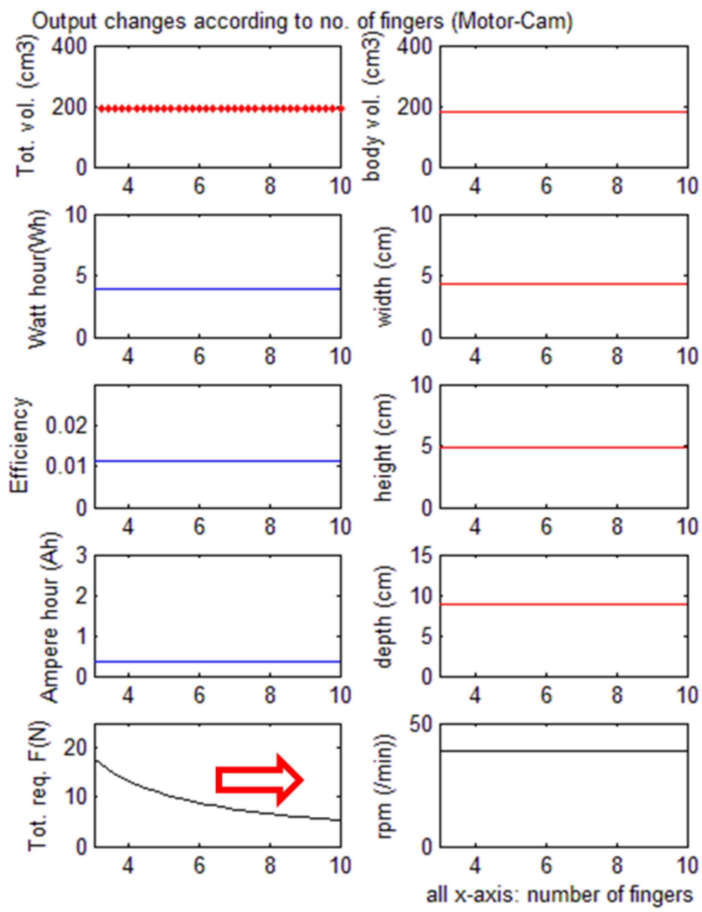


Figure 5-9. Output changes according to number of fingers (Motor-cam)

Figure 5-9 shows the output changes according to the change in the number of fingers. The only output variable affected is the total required force. It is because we are assuming that always same number of fingers (three in this case) are squeezing tube at an instance although the number of fingers change. When the number of fingers increases, each finger's width reduces, and it leads to smaller area of tube squeezed at one time. The figure suggests to increase number of fingers to actuate with a smaller actuator.

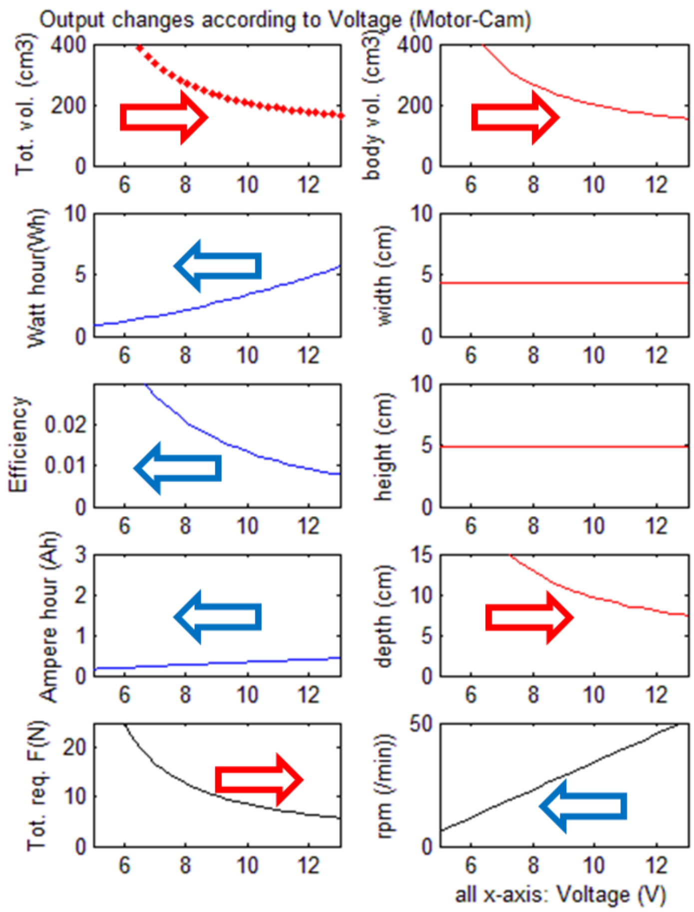


Figure 5-10. Output changes according to applied voltage (Motor-cam)

Figure 5-10 shows the output changes according to the change in the applied voltage. The voltage change affects most of the output variables. The reason that the width and height of the pump does not change is because it is mainly affected by the size of tube. However, the voltage affects the depth of the pump affecting the volume in the same pattern. The figure shows that the voltage should decrease to obtain a higher efficiency, but should increase to reduce the volume of the pump body and required actuator size by reducing total required force generation. However, the total volume that embeds power consumption and pump body volume in terms of size indicates that increasing the voltage is better. Thus, to reduce the total volume of the pump and required force generation, the voltage should increase.

5.6.2 Flexinol Finger Pump Model

This section will investigate the trends in a Flexinol driven finger pump. Many outputs will have the same trend as was in the motor-cam driven design, but it was checked again to avoid any confusion.

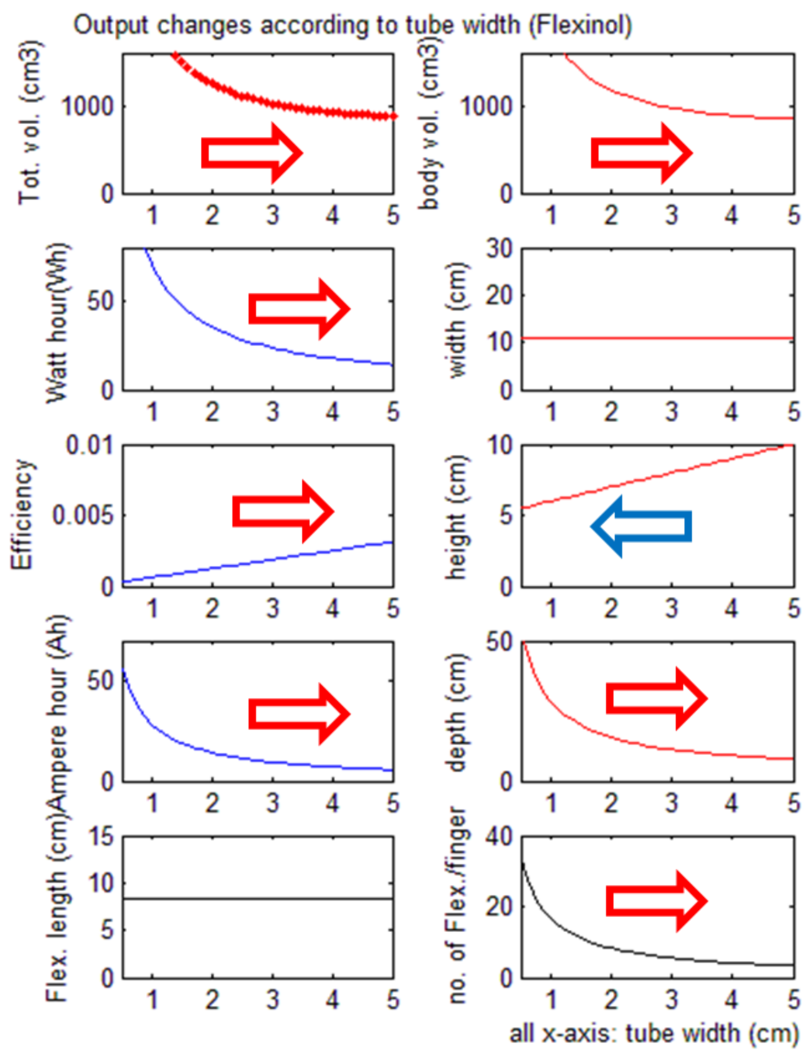


Figure 5-11. Output changes according to tube width (Flexinol)

Figure 5-11 shows the output changes according to the change in tube width. It suggests increasing the tube width since it will reduce the total volume of the pump while reducing number of Flexinols that should be connected to each finger.

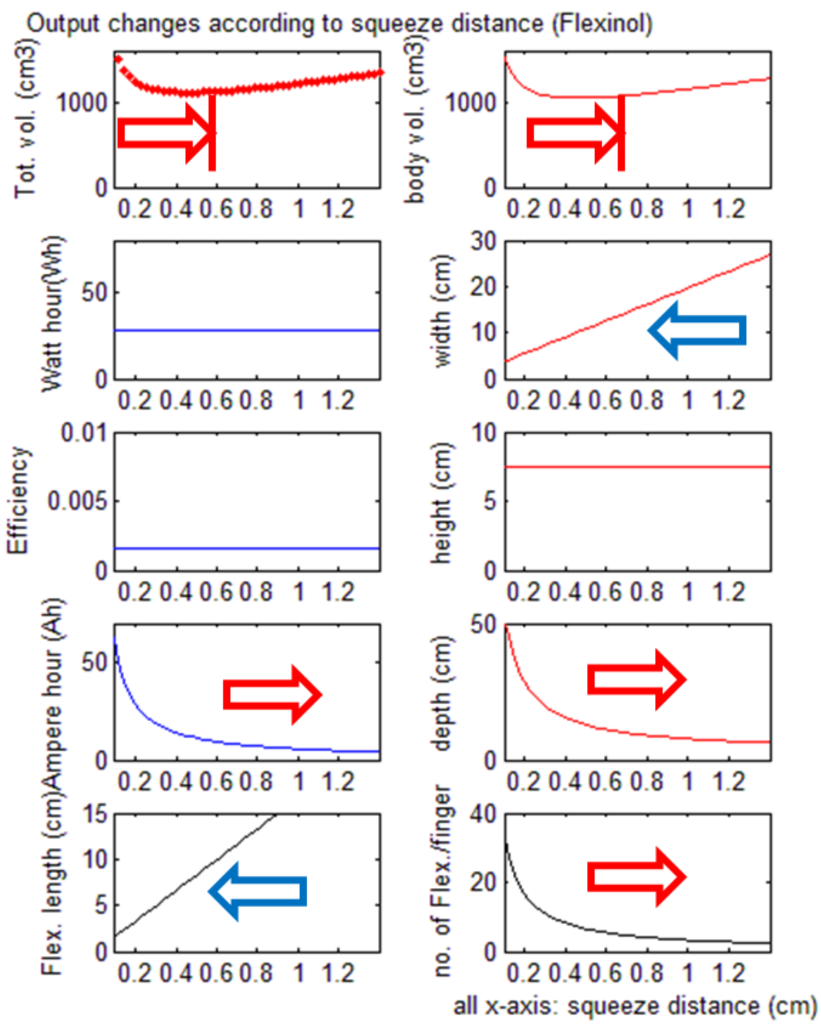


Figure 5-12. Output changes according to squeeze distance (Flexinol)

Figure 5-12 shows the output changes according to the change in squeeze distance. It does not affect the efficiency of the pump, but ampere hour decreases as the squeeze distance increases. To decrease the length of Flexinol and width of the pump, the squeeze distance should decrease. However, to reduce the number of Flexinols per finger and depth of the pump, the squeeze distance should increase. The total volume of the pump reduces as the squeeze distance increases up to a certain point, then starts to increase since the increase in the width of the pump starts to dominate. It suggests that a designer should compromise and find a middle range squeeze distance.

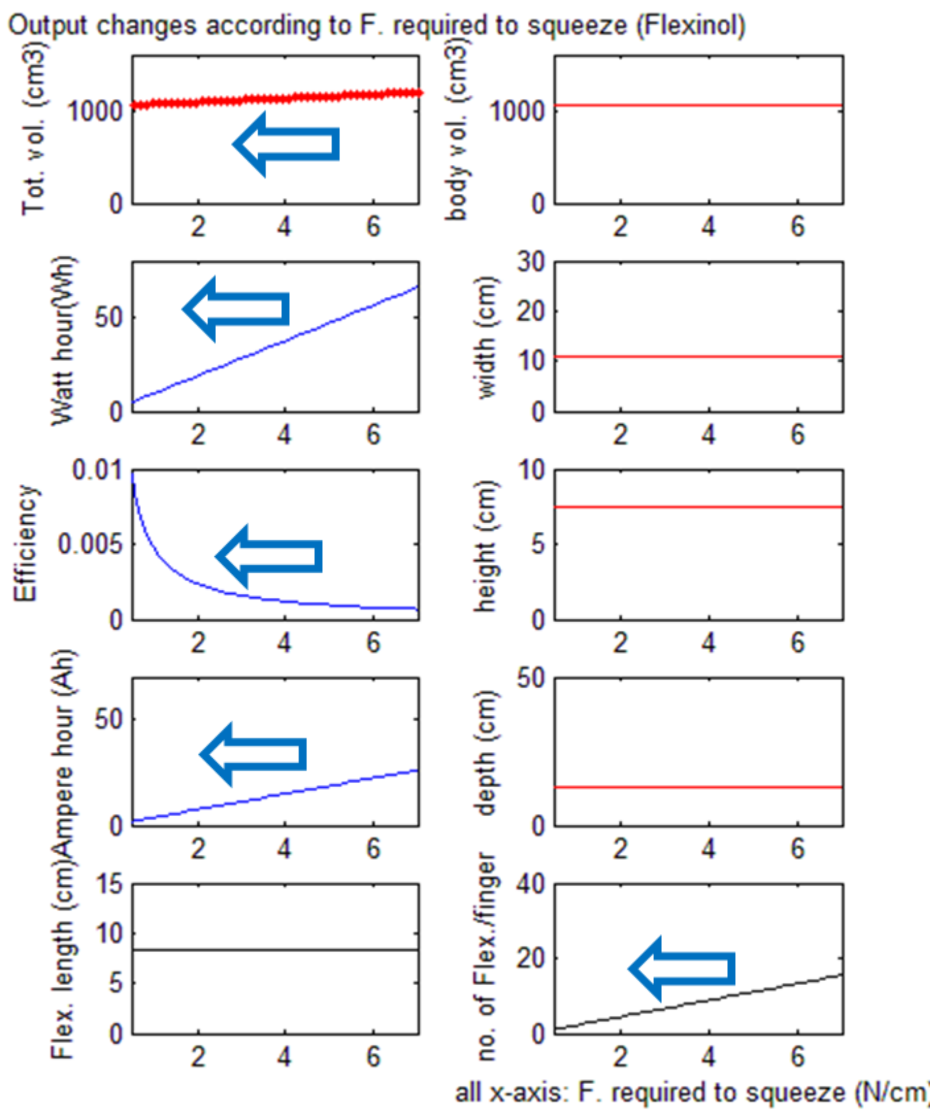


Figure 5-13. Output changes according to F required to squeeze (Flexinol)

Figure 5-13 shows the output changes according to the change in tube compliance. In this Flexinol driven pump model, the relationship between the required force and power consumption is embedded (motor-cam driven pump model did not embed this relationship).

It suggests reducing the force required to squeeze by using compliant tubes. It reduce the total volume of the pump and number of Flexinols attached to each finger.

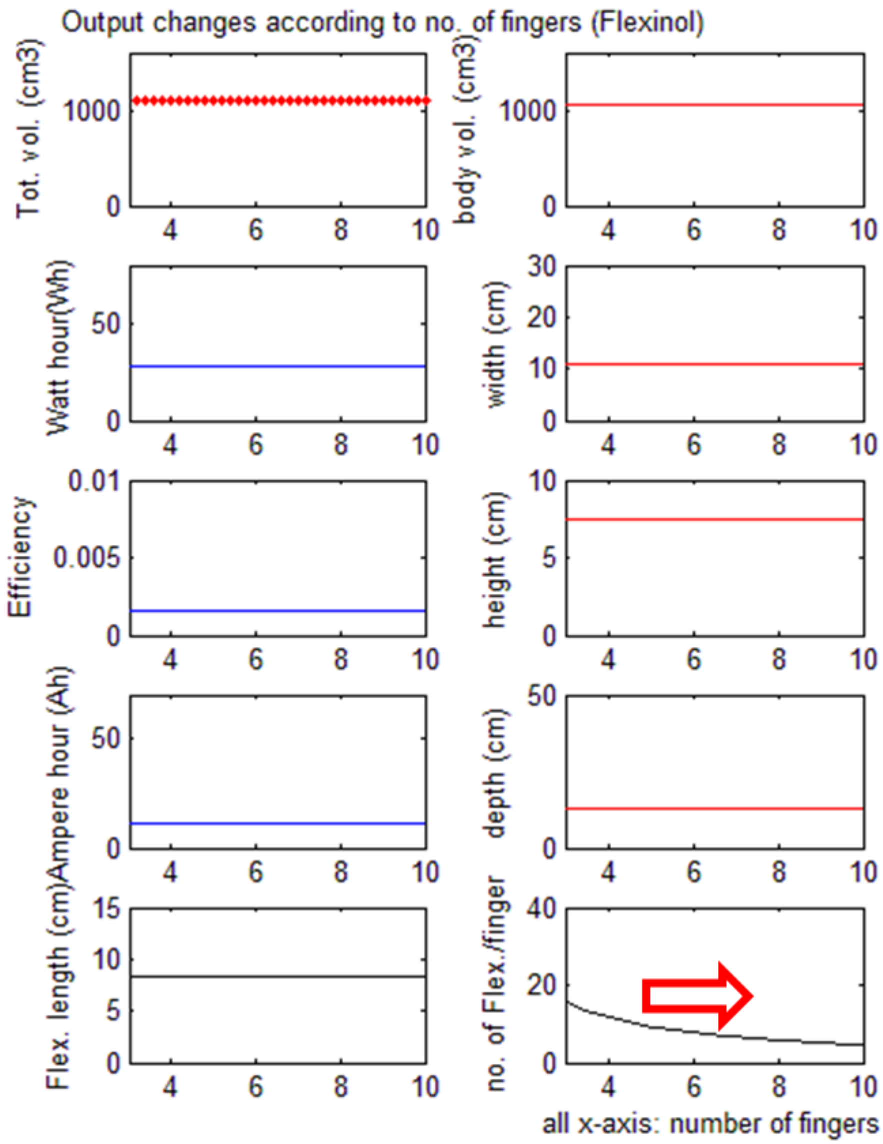


Figure 5-14. Output changes according to no. of fingers (Flexinol)

Figure 5-14 shows the output changes according to the change in number of fingers. It is desired to increase the number of fingers since it reduces the number of Flexinols attached to each finger while keeping all the other outputs the same.

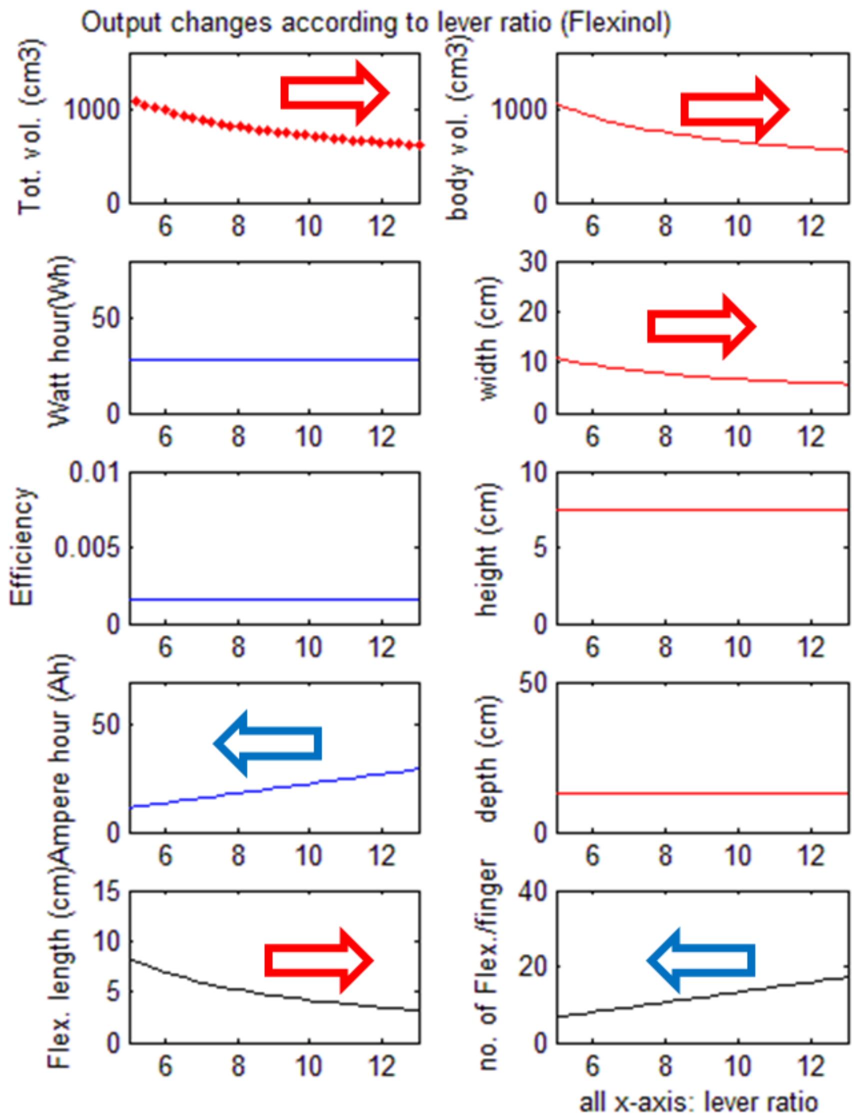


Figure 5-15. Output changes according to lever ratio (Flexinol)

Figure 5-15 shows the output changes according to the change in lever ratio. Increasing the lever ratio will reduce the pump volume by reducing the Flexinol length and pump width. However, it will increase the number of Flexinols attached to each finger. Thus, designer will have to find a middle value that yields small enough volume while keeping the number of Flexinols/finger reasonable so that actual manufacturing is possible.

5.7 Conclusion

Analytical pump models that include limited number of design parameters are created for both motor-cam driven pump and Flexinol driven pump. The experimental settings and/or conditions are expressed with a lumped constant, oval constant. The relationship between input and output variables is investigated and suggestions are shown in Table 5-4.

Table 5-4. Suggestion for optimizing the pump design

	Motor	Flexinol
tube width	increase up to a certain point	increase
squeeze distance	increase	increase up to a certain point
Force required to squeeze	reduce	
no. of fingers	increase	
Voltage	increase	N/A
lever ratio	N/A	increase up to a certain point

For both actuators, it is better to reduce the force required to squeeze. It matches the suggestion from Chapter 4 that suggested to use a very flexible tube, penrose drain tube.

The research question this chapter focused on was:

> *Can an analytical model that embeds relationships between limited number of design parameters (which includes design requirements, design specifications, and characteristics of an actuator) be reliably used to explore the design space under a predefined experimental setting?*

The creation of analytical pump models with limited number of design parameters with oval constant have allowed for an exploration of all the different parameters associated with pump design. The reliability of the model will be validated and the oval constant of the experimental setting will be found in Chapter 7. Then, the pump design that best fits a portable renal replacement system will be found in Chapter 8 using the pump model.

CHAPTER SIX

DETAIL DESIGN OF THE PUMP FOR A PORTABLE RENAL REPLACEMENT SYSTEM

Chapter 3 finalized the layout of the finger pump design and actuators that will be used for the pump. Chapter 6 provides the detail design of the pump. The goal of the design is minimizing the total pump size and weight of the pump. To achieve the goal, all the components are designed and laid out for the minimization of the waste space. Since two different actauators are final candidates, two different detail designs are provided. For each design, the overview of the design will be shown and explained followed by each component design.

6.1 Motor-Cam Driven Design

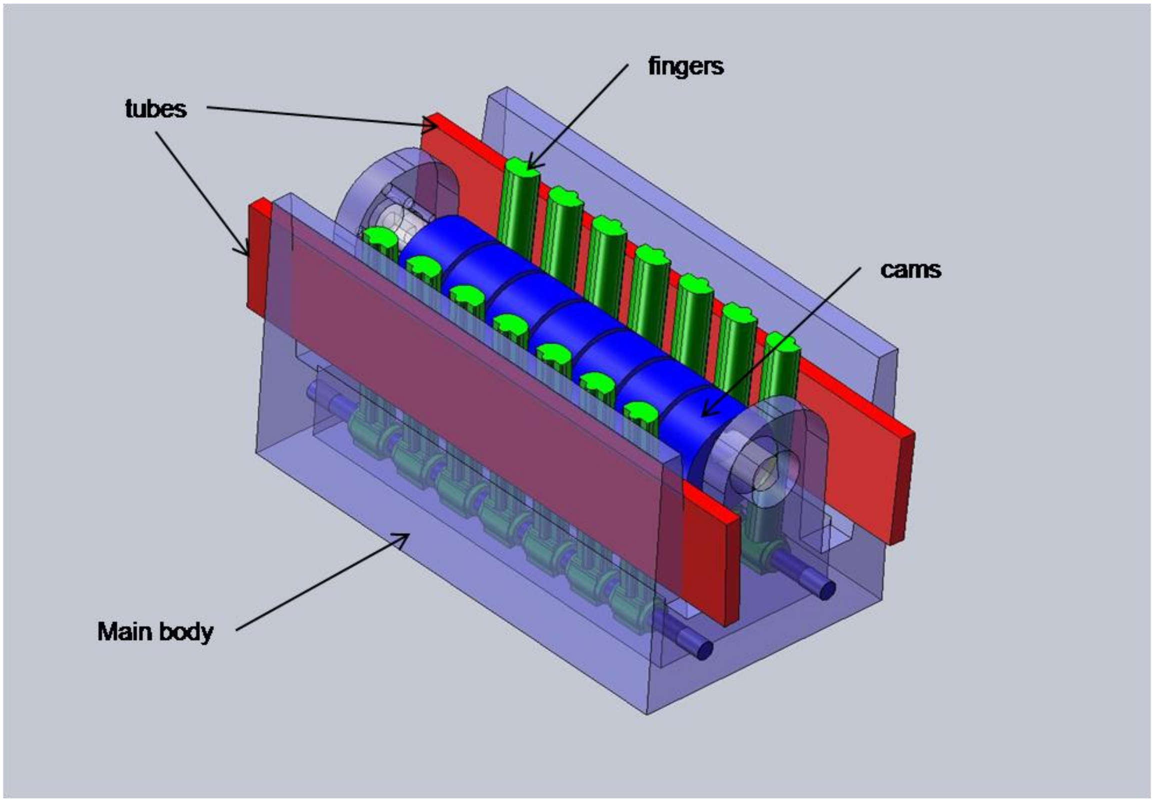


Figure 6-1. Overview of the motor-cam driven pump design I (suggestion)

Figure 6-1 shows the suggested motor cam driven pump design. Transparent light purple structure is the main body of the pump. Transparent light purple structure is the main body of the pump. The main body holds most components of the pump (cam shaft, motor shaft, gear shafts, finger shafts, and ball bearings at the end of cam shafts) and provides the plane surface for a tube to be compressed. Blue components in the middle are the cams. It is not clearly shown in the figure, but each cam is mounted on a shaft and oriented in different angles so that they can sequentially actuate fingers to compress the tubes. The red rectangular solids represent tubes that will be positioned between the pump body and fingers. Light green colored components are the fingers that will be positioned between cam and tube. Those are mounted on two shafts on each side and move back and forth as each cam actuates each finger. The fingers are positioned with some distance to maximize the efficiency as well as to reduce the friction loss. Although they are not shown in the figures, washers are mounted between fingers to keep the exact distance between them and reduce the friction loss by preventing direct contact between fingers. The motor is positioned under the camshaft, and gears will connect the motor shaft and camshaft to run together.

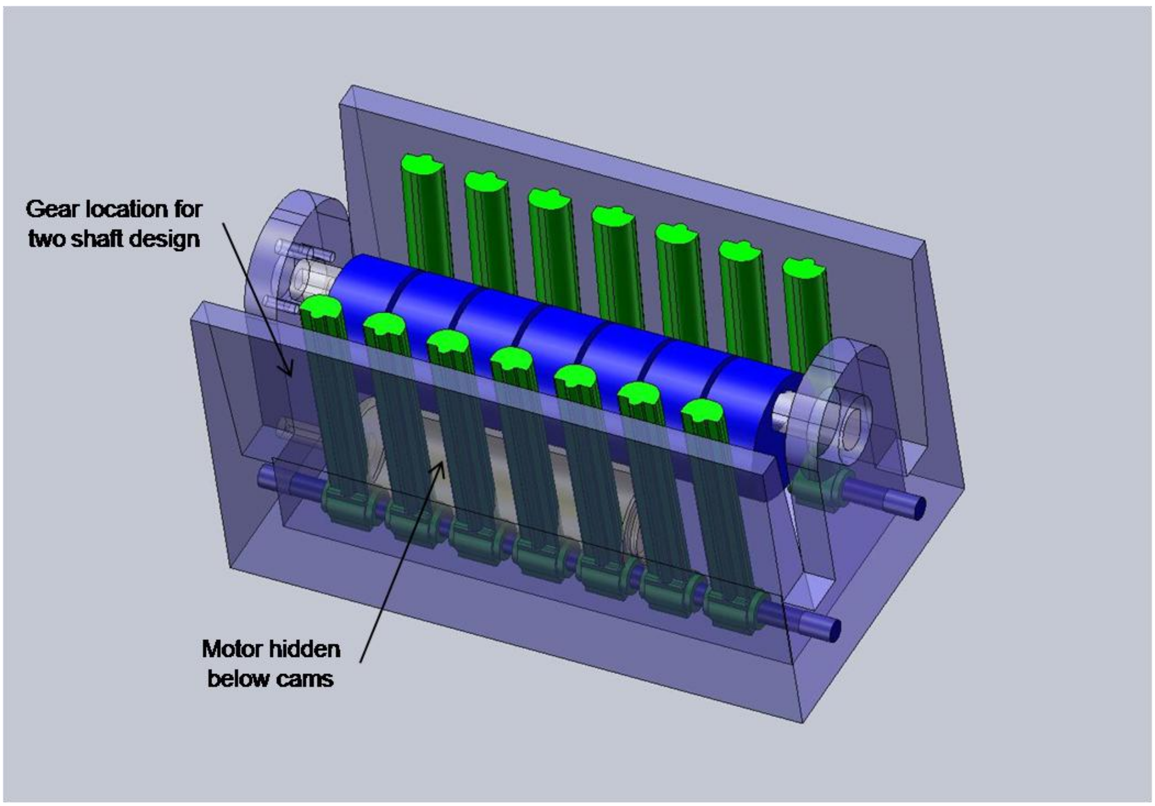


Figure 6-2. Overview of the motor-cam driven pump design II (suggestion)

Figure 6-2 shows the hidden motor location inside the main body under the cams in yellow color as well. However, it should be noted that the gear design that connects the two shafts (camshaft and motor shaft) is not fully provided but the layout is suggested as shown.

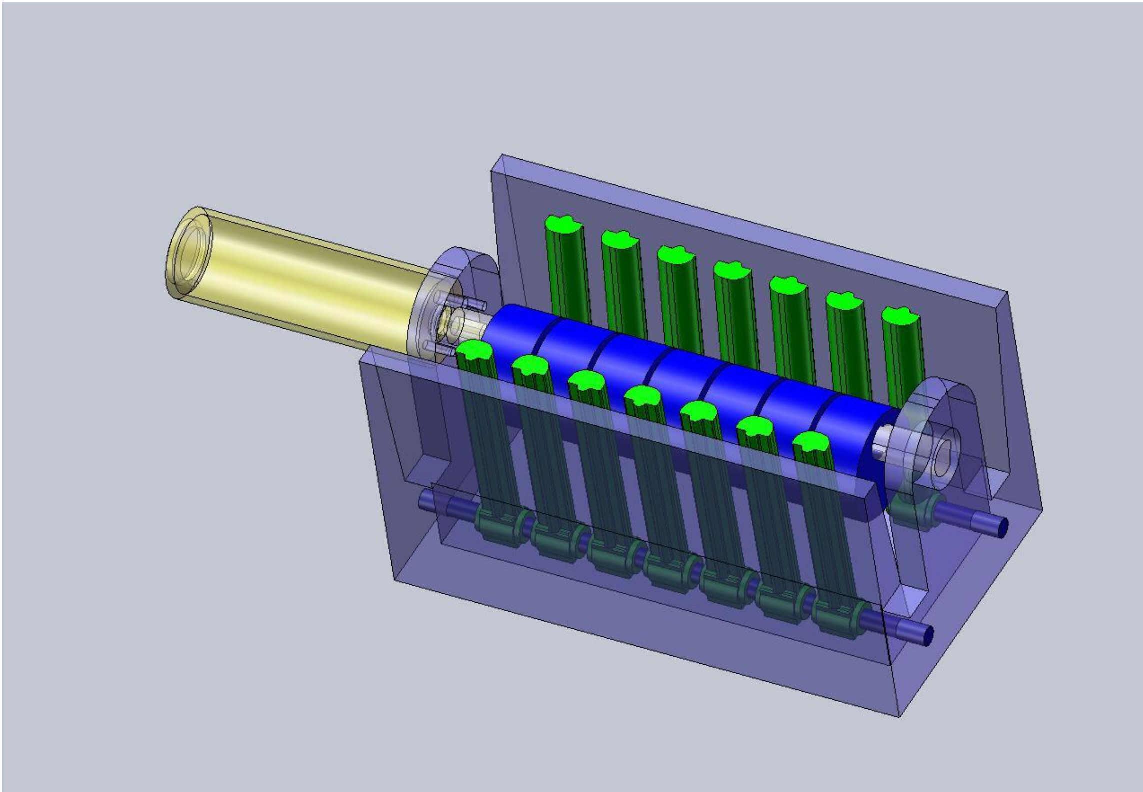


Figure 6-3. Overview of the motor-cam driven pump design III (built)

The pump design that was built and tested is shown in Figure 6-3. The design has exactly the same configuration as the suggested design for using a gear system to run two shafts (motor shaft and cam shaft) except that the motor is positioned outside the pump to directly connect the motor shaft to the camshaft without the use of a gear system.

6.1.1 Cam Design

Figure 6-4 shows the cam design. The cam is designed by cutting off some edges from a circle except for the part where complete compression is required. The compression start angle changes the orientation of the cam on the camshaft to change the compression starting point. By changing this compression angle, cams that compress the tube sequentially can easily be designed. Duration angle is the angle that the cam will be compressing the finger completely and holds at the position. As duration angle increases, the time that the tube is occluded will increase as well when the shaft rotation speed is kept the same.

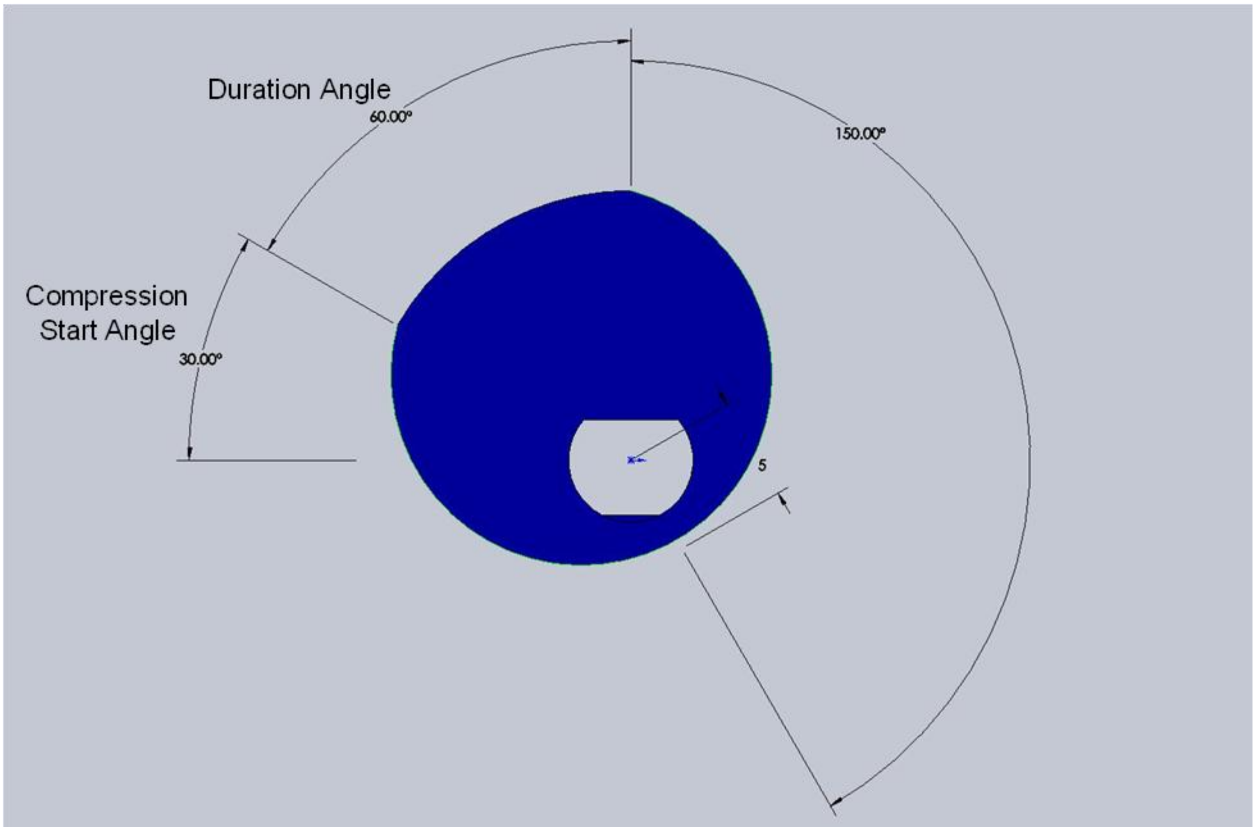


Figure 6-4. Cam design

This cam design has the advantage of enabling the complete control of tube occlusion time compared to an eccentric cam design that provides only one instance of occlusion. The design that was built and tested implemented seven cams that each actuate a finger on each side. Fiven middle cams had 60 degrees of duration angle while the first cam had 90 and the last finger had 180 degrees of duration angle. It was done to facilitate the flow by providing longer time for the tube to fill up after compression and reduce the backflow especially when the only finger stopping the back flow is the first finger.

6.1.2 Finger Design

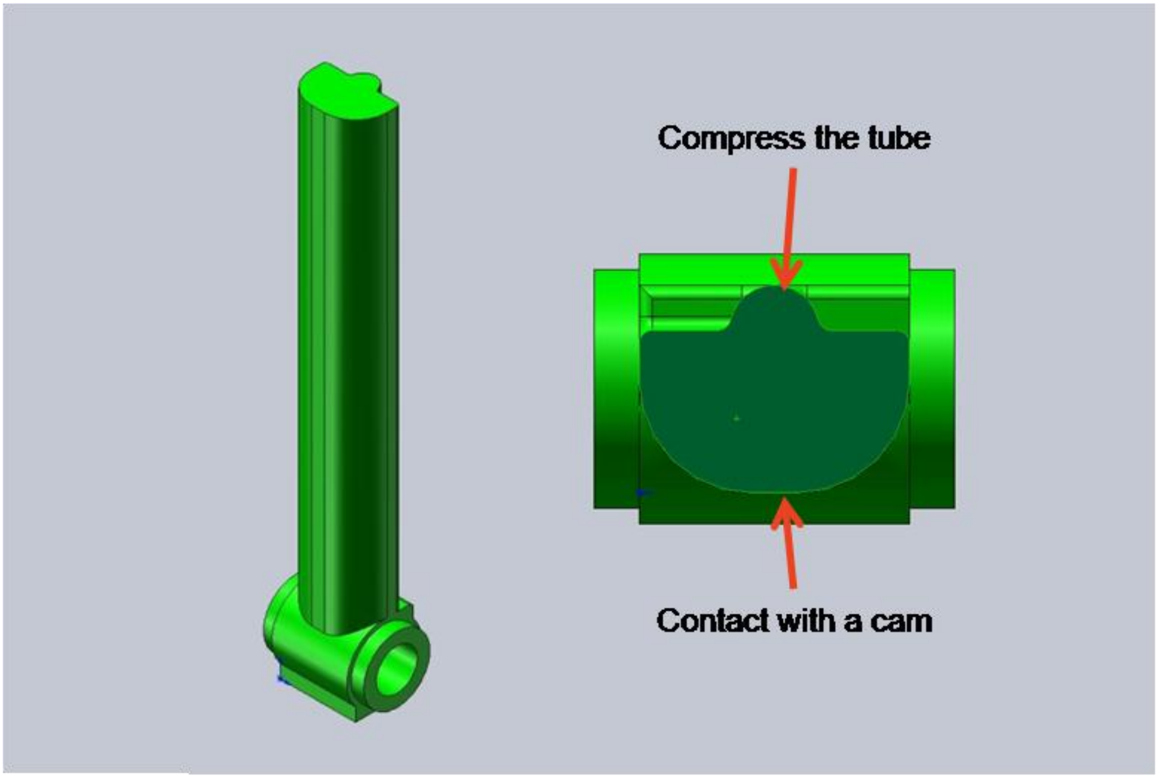


Figure 6-5. Finger design

Figure 6-5 shows the overview and top view of the finger design. The finger has a cylindrical hole to be mounted on a shaft and the thicker design around the hole than upper part of the finger to allow the least friction loss between a finger and the washer next to it (it could be rounded to provide line contact to reduce the friction loss even more). One important part to be noticed in this design is the reduced contact area between cam and finger, and finger and tube. Theoretically, the contact area between cam and finger should not affect the force required to compress, but in reality, it does. Thus, reducing the contact area with a cam will help reduce the energy consumption by reducing the friction loss. The reduced contact area between finger and tube induces the same effect by reducing the complete occlusion area. As mentioned in Chapter 4, it will not affect the flow rate of the pump as long as the tube is completely occluded by the contact area and not too much of fluid is kept in between contact areas of fingers. In addition, the small occlusion area will reduce any possible blood cell damage.

6.2 Flexinol Driven Design

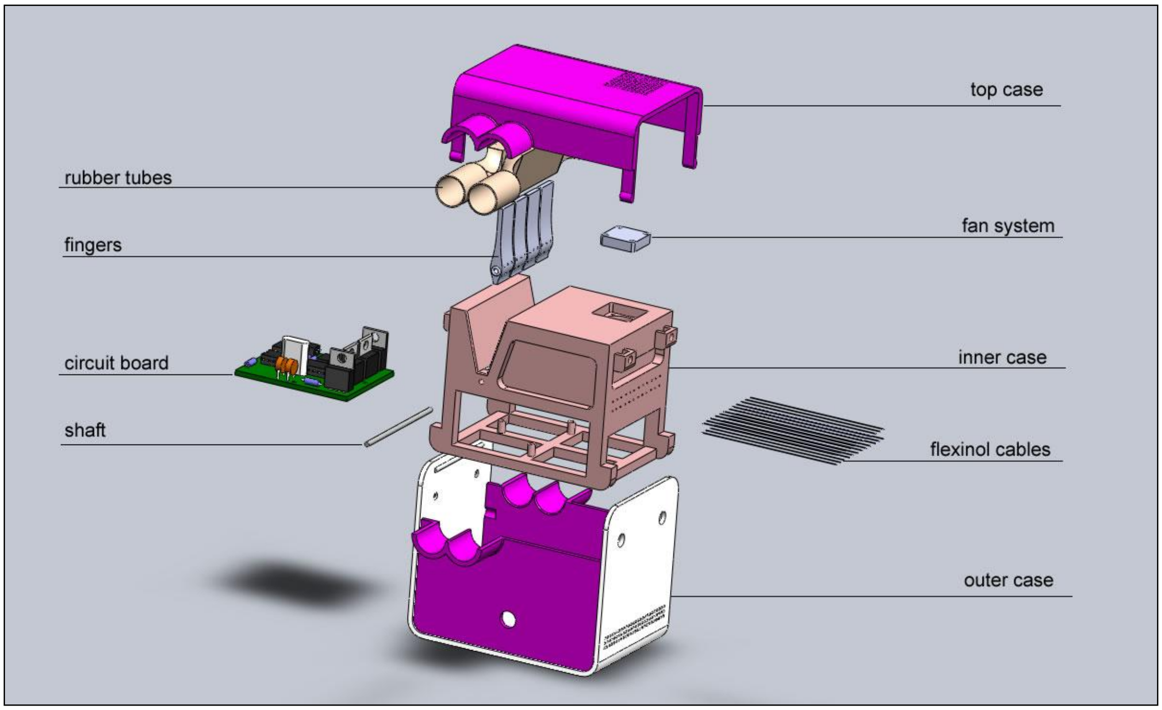


Figure 6-6. Assembly exploded view of the Flexinol driven pump design

Figure 6-6 shows the assembly exploded view of the Flexinol driven pump. Although the layout in Figure 3-9 shows Flexinols positioned on both sides, Flexinols are positioned on one side of the pump in this detail design to reduce the size of the pump. The main body supports and connects the main working components (finger shaft, Flexinols, fan system, and circuit board).

The circuit board, fan system, and case outside the main body are required due to the characteristic of Flexinol. The circuit board is required to provide appropriate signal to

Flexinols since continuous DC supplied from a battery cannot be directly used to actuate Flexinols sequentially. Fan system is required since Flexinols are heated when those are contracting. If the ventialation fails, the whole pump could fail due to overheating. A case outside the main body is required to protect the inner components and the user against heat burns and electrical shock from the Flexinols. The top case is required to allow daily tube replacements and occasional cleaning of the fan.

The next few sections provide the reasons for the shape of each component.

6.2.1 Fingers and Flexinol

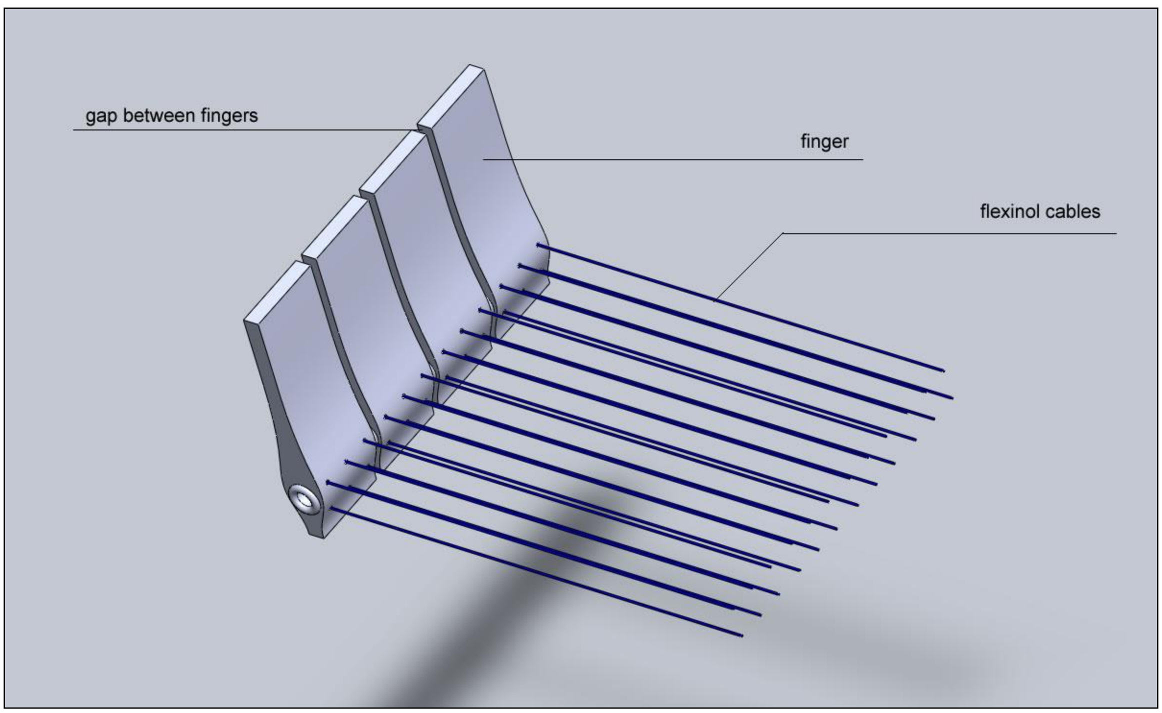


Figure 6-7. Fingers and Flexinols

The fingers compress tubes on either side of their axial plane. Hence, they should be symmetric about this plane as shown in Figure 6-7. The figure also shows how fingers and Flexinols are connected. The fingers are the only components that move during the normal operation of the device as they are actuated by the Flexinols. Thus, the fingers have to be designed to sustain the force exerted by the Flexinols and amplify the contraction of the Flexinols to achieve the required compression of the tubes while reducing the size of the pump. Since the fingers should amplify the contraction of the Flexinols, those are designed as levers with Flexinols close to the pivot point. Fingers are mounted on a shaft as was in the motor-cam driven design. One set of Flexinols is connected above the hole, and other set of Flexinols is connected below the hole to provide tube compression on the both sides with only contraction of Flexinols. Looking at the finger from the right side as shown in the figure, the finger will be pulled when the upper side Flexinols are contracting. Then, the bottom side Flexinols will contract to pull the finger back to the other side (push away from this side). Since Flexinols exert concentrated force close to the shaft axis, the region close to the hole for the shaft is thickened to increase its load bearing characteristics. The rest of the finger is parallel to the surface of the tube channel. Semi circular spacer rings have been added to maintain a gap between them as the finger design for the motor-cam driven design. They reduce friction between fingers during the rotation by minimizing the contact area between the fingers almost to a line.

Although two different finger designs are suggested for each pump, it does not mean that these designs are applicable only for each pump. The features of the fingers can be mixed together to improve the finger's functionality.

6.2.2 Main Body Design

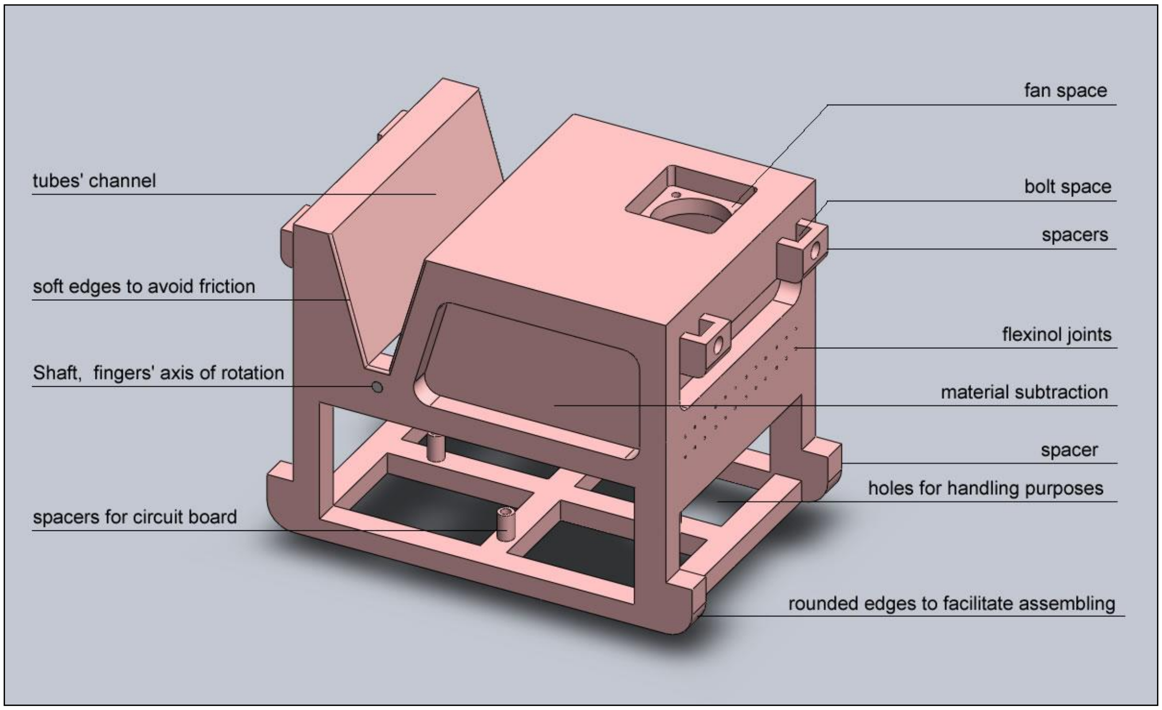


Figure 6-8.Main body of the Flexinol driven pump

The main body shown in Figure 6-8 positions and supports all the functional components, including tubes, fan system, circuit board, and Flexinols. The main body has the main tube channel, a slot for fan, spacers, and the pattern of holes for holding Flexinols. Pockets are created on the bottom and side walls to provide space for handling fingers, Flexinols, and circuit board installation, and reduce the weight of the part. The trapezoidal tube channel provides the surface against which the fingers compress the rubber tubes. Its edges are rounded to avoid puncturing the tubes. In the upper top face of the main body, a slot is created to place the fan. Placing the fan on the top surface allows access for cleaning the fan periodically without total disassembly of the pump. Open gaps are avoided in the top area to avoid user access to the Flexinols and circuit area. A series of cylindrical supports position the circuit board at the bottom of the main body. Spacers are provided on the outer surface of the main body to maintain a gap between the main body and the case. This gap allows space for electrical connections with the Felxinol and also maintains a gap for air flow between the main body and the outer case. The rounded edges on the bottom spaces eases the insertion of the main body in the case.

6.2.3 Case Design

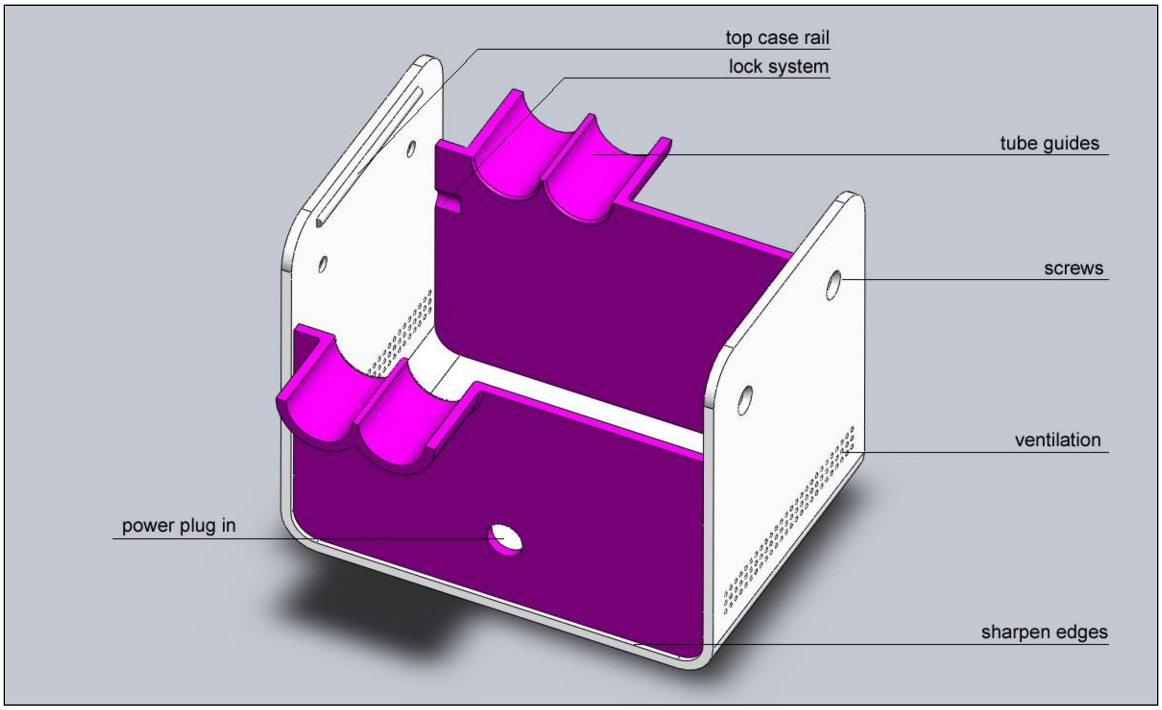


Figure 6-9. Outer case

The case shown in Figure 6-9 protects the user from contacting the internal components and facilitate ventilation. Once the main body fits into the outer case, four screws assure its position. Sharpened edges provide the necessary traction to hold it, while detaching the top cover. The rigid cylindrical tube ends are clamped between the cylindrical tube guides in the outer case and case top. Cooling air comes in through two patterns of holes in the bottom of the case to reduce the cooling time of the Flexinol and avoid overheating of the device.

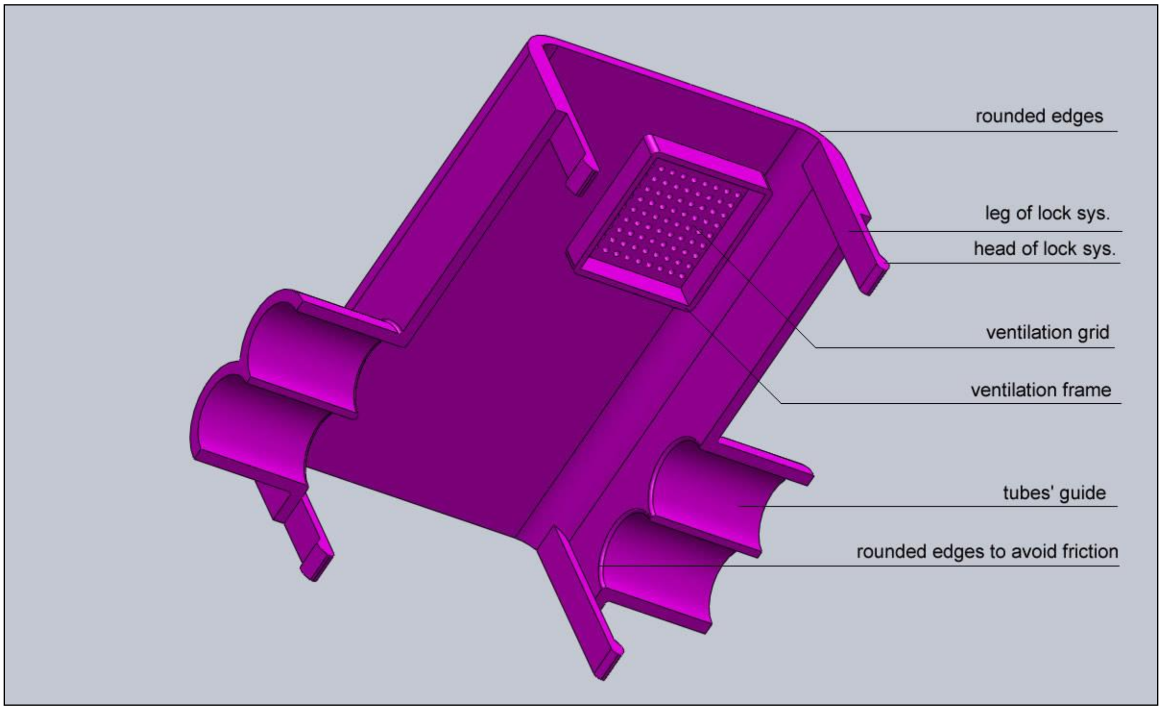


Figure 6-10. Top case

Figure 6-10 shows the top case that positions and locks the rigid cylindrical end sections of the tube, facilitate ventilation, and allow daily replacement of tubes. The tubes' guides complement those from the outer case to hold the rubber tubes in position while the pump is in function. The top ventilation grid and the ventilation frame contribute to the cooling process. The grid allows heat extraction from inside, and the frame channels the air flow. Such channeling is required because of the gap between the top case and the main body. With no such frame, the fan system could be recirculating heat rather than actually extracting it. Finally, the lock system is based on a flexible leg with a head at one end which locks into a slot in the outer case. Pressing the top case from the side retracts the head from the slot of the outer case and allows the pump to be opened.

6.3 Tube Plate Design

As mentioned in Chapter 4, engineering tolerance could be critical for a finger pump. Smalle changes in the tube thickness could bring the difference between complete and uncomplete occlusion of the tube. If the tube plate is set to a minimum thickness of the tube so that the tube is always completely squeezed even in case the tube is thin, the complete occlution can be guaranteed anytime. However, the actuator load will increase when the tube is too thick and requires much more force to completely squeeze the tube compared to a thin tube. To resolve these problems, this pump design added a rubber plate on the main body, so that the actuator load can be reduced when the tube is too thick and completely squeeze the tube all the time.

6.4 Tube Connector Design

This design proposes using a thin-walled flexible tube (penrose drain tube) to reduce the force required to compress it. However, the flexibility is required only for the part that is inserted in the pump. For the other part in the fluid circuit, using a tube commonly used for IV pole is a better choice since using flexible tube for the whole circuit could cause too much flow rate variation. In addition, the size of the cross section of the tube may vary depending on the requirements. Thus, to make the process easier for the experiment, changing the connector and tube and keeping the IV pole tube for other parts rather than changing the tube for a whole circuit is preferred.

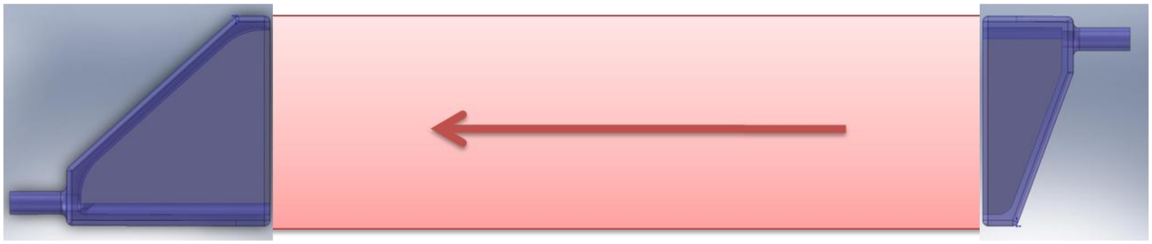


Figure 6-11. Tube connector design

Figure 6-11 shows an example of the tube connector design. The purple colored components at both ends are the connectors, and the red component in the middle represents the flexible tubing that will be inserted in the pump. The figure shows the configuration for the flow from right to left. Connectors have different design for each side to accommodate the flow by making the backflow hard once fluid comes into the tube. [19] For the blood flow, the blood drawn out from a patient's body through IV pole tube flows into the flexible tubing through the connector on the right. Then the blood will flow out of the flexible tube through the connector on the left and go back into the patient's body through IV pole tube.

The connector design may be slightly different depending on the size of the flexible tube. However, the main feature that connects the IV pole tube from the top side for the inlet and bottom side for outlet will remain the same.

6.5 Conclusion

This chapter suggested detail pump designs for both of Flexinol and motor-cam driven pumps. These designs are subject to change in dimensions or number of components, such as cam or finger, but the layout will not change. The design parameters such as number of fingers and finger width are determined using the pump model created in the previous chapter. Values that are believed to be close to the values for the best design are chosen.

The research question this chapter focused on was:

> *Can a pump that is more compact, lighter, and more energy efficient than currently existing pumps for the portable renal replacement system be designed with lower manufacturing cost, less noise, and less possibility of clogging?*

The detail designs were proposed based on the layout proposed in Chapter 3. In the following chapters, these pump designs will be validated for its performance by running experiments. When the final pump design for a portable renal replacement system is proposed in Chapter 8, the hypothesis that adopting a different working principle and a different actuator will allow achieving the goal listed in the research question will be answered.

CHAPTER SEVEN

VALIDATION

Detail designs for both actuators are provided in Chapter 6. The next step is to validate the pump model created in Chapter 5 using the pumps designed in Chapter 6 so that the model can be used for the design space exploration and optimization in the following chapter. This chapter will go through the validation of the model with water experiment first. Then, *in vitro* experiments will be conducted to prove the the validity of the pump use for the portable renal replacement system.

7.1 Validation of the Pump Model

This section will validate the pump model by conducting water experiment with the pumps designed in Chapter 6. It will prove that the model can provide the expected pump performance for both designs.

Many common concepts such as the effect of cross section area, fluid viscosity, and pump head change are tested with motor-cam design. The effect will be the same for both Flexinol driven pump and motor cam driven pump. After validating the pump model for the common factors, experiments with rpm change were conducted using both motor-cam driven pump and Flexinol driven pump to find the oval constant and prove that the pump model can match the Flexinol design as well.

7.1.1 Effect of Cross Section Area Change on the Flow Rate

The laboratory setup is shown in the top picture of Figure 7-1. The pump was operated by the DC power supply. Pen rose drain tube is used to reduce the force required to squeeze.

The water flowed from the water reservoir on the left to the water disposal on the right through the pump in the middle. As explained before, the pump was located at the lowest point in the circuit to help the shape recovery of the flexible tube after each compression.

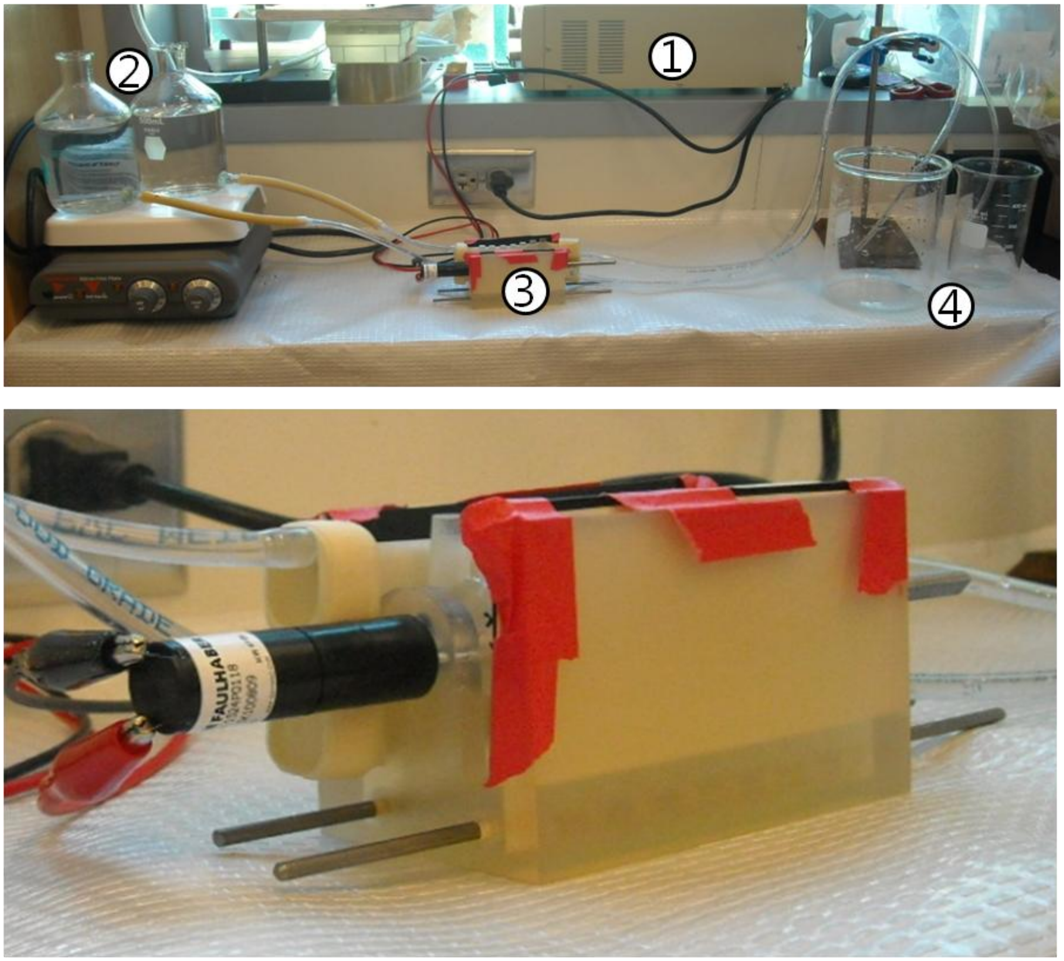


Figure 7-1. [Experiment I] The laboratory setup for water experiment with motor-cam pump. (Two small tubes on one side) ①the DC power supply, ②the water supply reservoir, ③the motor cam driven pump, ④the water disposal

To test the effect of the cross section area difference, two small tubes were located on one side as shown in the bottom picture of Figure 7-1. One of the reasons that two small tubes than one large tube were used is the different flexibility of the small and large tubes available in the market. Larger tubes tend to be more flexible than the smaller one, and it led to inconsistent flow rate. Other reason is to test the effect of changing the cross section area for the same size tubes. Although the two tubes are the same ones, due to the geometry of the pump, the lower tube happens to have smaller cross section area than the top tube. The experiment results are shown in Table 7-1.

Table 7-1. [Experiment I] Water experiment results with two small tubes on one side

Voltage (V)	10.8
rpm	38.3
top tube (ml/min)	145
bottom tube (ml/min)	94

As expected, the top tube that has larger cross section area yielded much higher flow rate than the bottom tube. The top tube cross section area was approximately 1.586 times bigger than the bottom tube (top: 14×8.5mm, bottom:15×5mm), and the flow rate difference (1.54:1) closely matched the ratio proving that the cross section area increase does increase the flow rate linearly. The result also shows that using a bigger tube can generate the flow rate of 240ml/min in the given setting.

An additional experiment was conducted to test the same effect under slightly different condition. This time, the pump operated with a tube on each side as shown in Figure 7-2 to prove that the pump will perform the same way when both sides are pumping. In addition, the water disposals were located higher than the previous experiment to increase the pump head and prove that the cross section area ratio will affect the flow rate the same way under different setup.

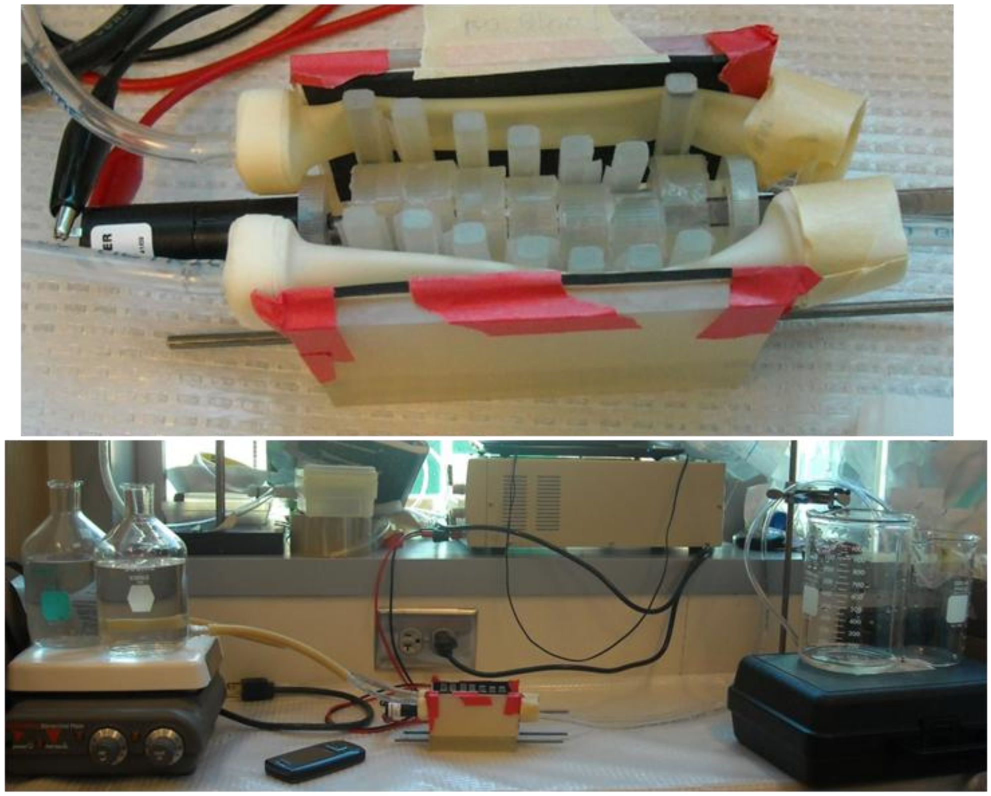


Figure 7-2. [Experiment II] The laboratory setup for water experiment with motor-cam pump. (Two small tubes on one side)

Two different cases were tested, one with the tubes in the middle of the fingers, and other with the tubes on the top as the top tube in the previous experiment. Table 7-2 shows the experiment results. The different rpm from the previous experiment for the same voltage can be noticed. The reason for the difference is because the plate against which the tube is compressed is thickened by adding a paper ensures complete occlusion of the tube. As explained before, complete occlusion is very important for reducing the backflow.

Table 7-2. [Experiment II] Water experiment results with a tube on each side

a tube on each side (middle)		a tube on each side (top)	
Voltage (V)	10.8	**Voltage (V)**	10.8
rpm	31.60112	**rpm**	31.60112
back (ml/min)	100	**back (ml/min)**	130
front (ml/min)	100	**front (ml/min)**	145

The ratio of two cross section areas on the top and the middle were about 1.25 (top: 14×8.5mm, middle:14×6.8mm), and the result matches the ratio (1.3). In addition, even with the increased head and lower rpm, the flow rate of the experiment with the tube on the top provides almost the same flow rate as the previous experiment. It shows that the previous experiment did not occlude the tube completely, and by completely occluding the tube in this experiment, the flow rate was kept the same under a more difficult condition. The different back side and front side tube flow rates are due to slightly different thickness of the pump plate. However, this difference can be controlled and is useful for providing different flow rate for each side.

7.1.2 Effect of Supply Side Head Change on the Flow Rate

This pump requires the supply side head to fill the tube because flexible tubes are used. In addition, the portable system could implement several dialysate bags to change every 30 minutes or so. Thus, the pump should be able to generate stable flow rate during the treatment whether the dialysate bag is full or emptied. Below experiments were conducted to verify the effect of supply side head change on the flow rate.

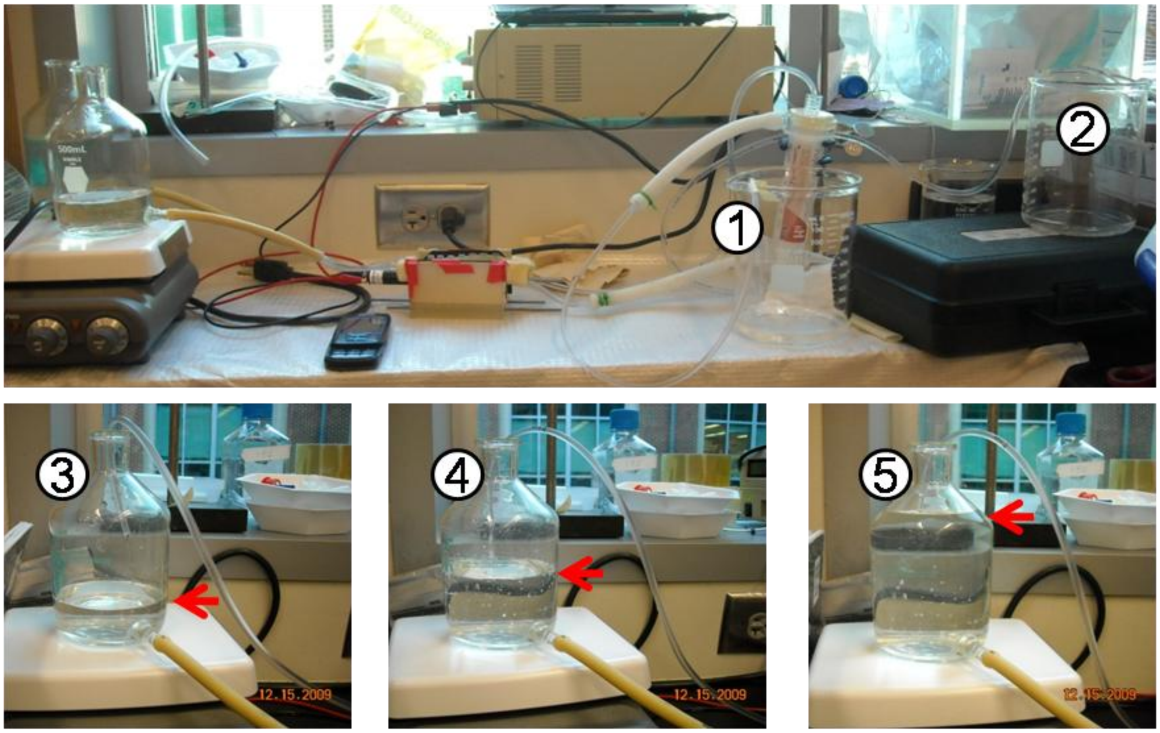


Figure 7-3. [Experiment III] The laboratory setup for water experiment with motor-cam pump. (with different supply side head)

Figure 7-3 shows the experiment setup. The flow circuit included a dialyzer to provide the same setting as a portable renal replacement system. ① in the top picture is the disposal beaker for dialysate side, and ② is the disposal beaker for blood side. Both in *in vitro* experiment or in treatment, the blood goes back into the original jar or patient's body, so no disposal beaker is required. However, in this experiment, it is used for the flow amount measurement. Since the blood will go back into a patient's body in a real treatment or into the jar in *in vitro* experiment, the height of the beaker is set to that of the blood reservoir. However, the location of dialysate disposal can change depending on the circuit configuration.

The head difference coming from the dialysate reservoir was tested as shown in the bottom picture of Figure 7-3. Four different water levels were tested by starting from a full reservoir (about 550ml) and running the pump for 1 minute each time. To provide different water level in the reservoir, the dialysate side reservoir is not filled up while the blood reservoir always started from a full reservoir. Table 7-3 shows the experiment results.

Table 7-3. [Experiment III] Supply side head change vs. Flow rate

Voltage (V)	10.5			
rpm	44			
Blood side ② (ml/min)	95	95	100	97
Dialysate side ① (ml/min)	132	127	125	125

The experiment result with different dialysate reservoir level shows not too much of flow rate difference although the flow rate did reduce a little. However, it is negligible amount of change considering that the blood side flow rate also fluctuated a little although it always started from a full reservoir. It can be considered as a measurement error since the pump generates not a continous flow but a pulsatile flow. Depending on the starting finger that compresses the tube for the very first time in each run, the flow rate can vary. These result proves that in the given system configuration, the head difference in the jar is negligible. However, this test was limited to the head change in the jar, and the flow rate is still subject to a larger supply side head change.

The different flow rate for the blood side and dialysate side comes from the different height of the disposal side as shown in the figure. More indepth exploration of the effect of different disposal side head change on the flow rate follows in the next section.

7.1.3 Effect of Disposal Side Head Change on the Flow Rate

Effect of disposal side head change on the flow rate was tested in in vitro experiment setting as shown in Figure 7-4.

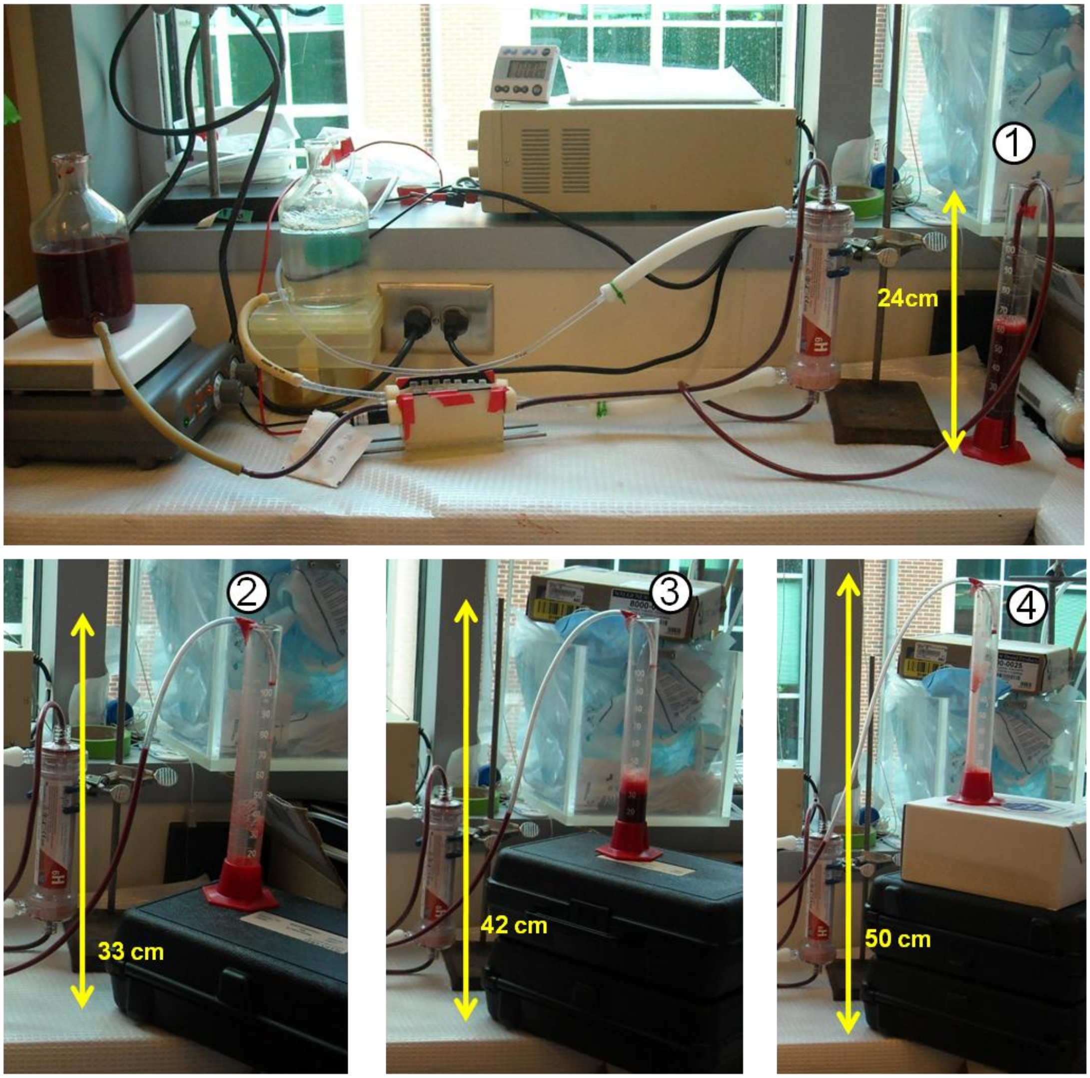


Figure 7-4. [Experiment IV] The laboratory setup for *in vitro* experiment with motor-cam pump. (with different disposal side head)

The blood flow rate was tested while both dialysate and blood were running through the dialyzer. Different disposal side head was tested for four different heads with different rpm for each configuration. The test results are recorded in Table 7-4.

Table 7-4. [Experiment IV] Disposal side head change vs. Flow rate

	voltage (V)	10.8	9.4	8
head (cm)	rpm	42.46285	34.18803	26.58789
24		96	90	68
33		71	67	49
42		54	41	32
50		38	22	-

The test results for 50cm head with 27 rpm is not provided because the fluid was not able to climb up the height with the slow flow rate. The results are plotted in Figure 7-5.

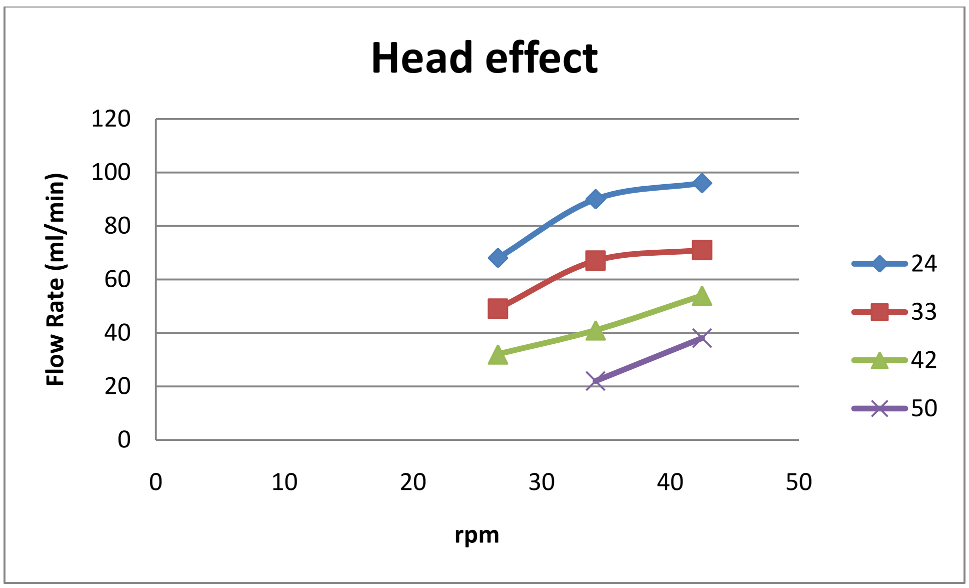


Figure 7-5. [Experiment IV] Disposal side head change vs. Flow rate

The figure shows that the flow rate is affected a lot by the disposal side head difference. The lower disposal side head yielded higher flow rates than higher head while increasing with increased rpm. It shows that the disposal side head should be seriously considered in designing a portable renal replacement system configuration. In addition, it shows that the oval constant will change a lot according to the head effect. Since the oval constant can be determined according to the flow rate change when all the other design variables are kept the same, it shows that the oval constant for 24cm height will be about 4 times larger than that for 50cm (90ml/min:22ml/min).

7.1.4 Effect of Fluid Viscosity Change on the Flow Rate

This pump is designed to pump two different fluids at the same time. For the use in a portable renal replacement system the blood and the dialysate are the two fluids. However, those two fluids have different viscosity. To make sure that two different fluids can be pumped appropriately, the effects of different fluid viscosity of the flow rate is tested.

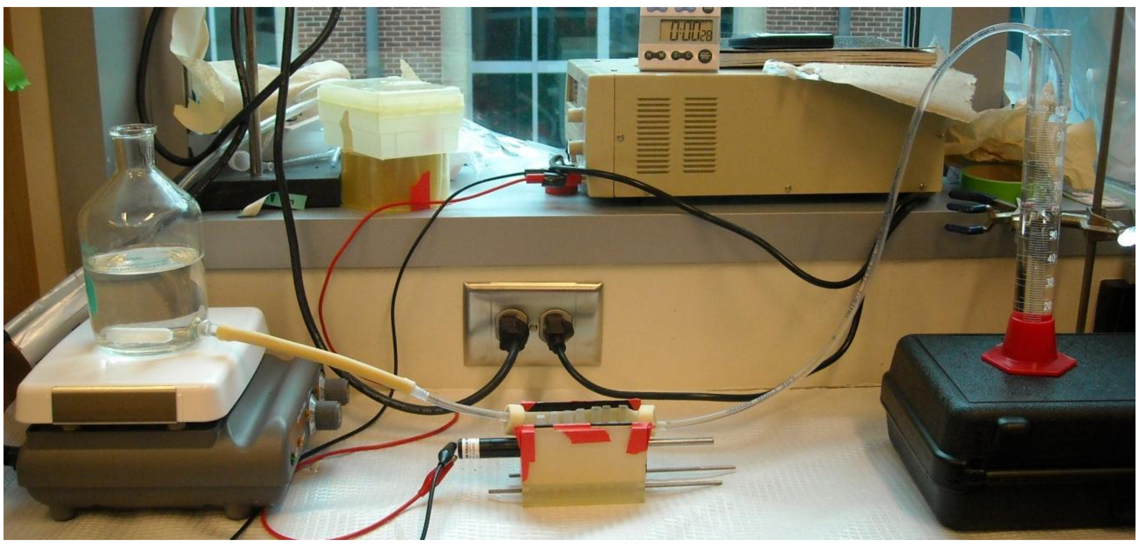


Figure 7-6. [Experiment V] The laboratory setup for water experiment with motor-cam pump. (fluid viscosity)

Figure 7-6 shows the experiment setup for the viscosity test. The fluid flowed from the jar on the left to the cylinder on the right. To control the viscosity of the fluid, glycerin was added to the water. Table 7-5 shows the flow rates for different fluid viscosities.

Table 7-5. [Experiment V] Fluid viscosity vs. Flow rate

Glycerin percent weight(%)	Viscosity (mPa s)	Flow rate (ml/min)			
12.6	1.4	132	132	132	132
22.4	1.9	136	136	136	136
30.2	2.5	134	134	134	136
36.6	3.3	138	138	-	-
41.9	4.1	134	132	134	-
52.0	6.9	136	136	136	138

As shown in the table, the flow rate almost had no difference for different viscosity. Considering that different viscosities up to 7 mPa s had no difference in the flow rate, the flow rate will be the same for both the blood (1.84 mPa s) and water (1.005 mPa s). It also shows that the oval constant can be kept the same when fluid viscosity is in the tested range.

7.1.5 Effect of RPM Change on the Flow Rate

Motor cam driven pump was used to prove that the flow rate of the pump linearly increases as rpm increases. Figure 7-7 shows the laboratory setup for the experiment.

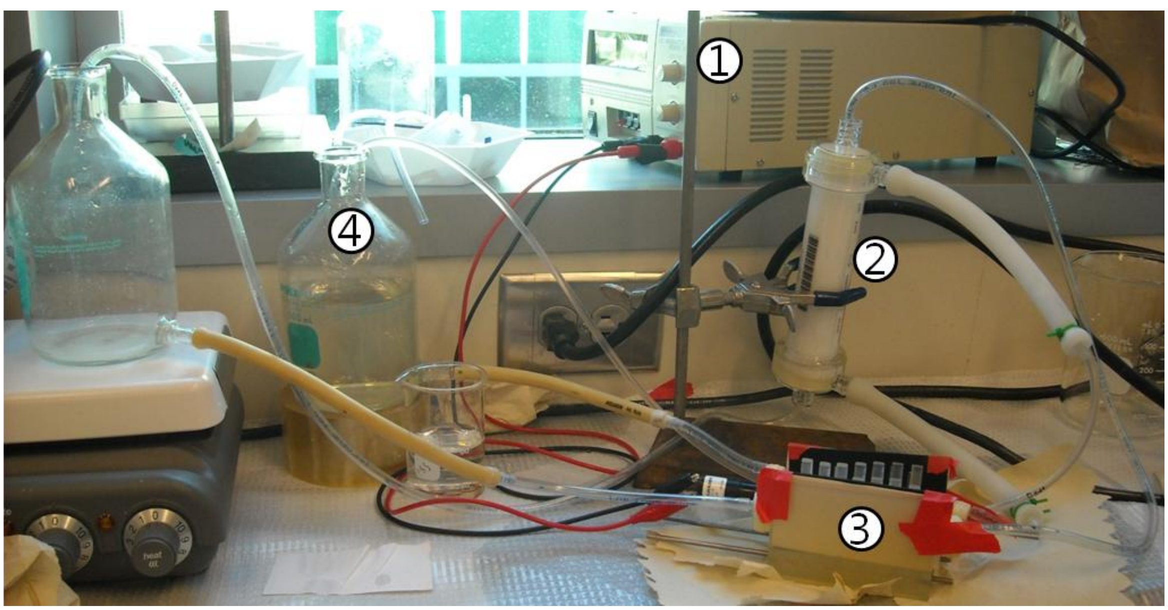


Figure 7-7. [Experiment VI] The laboratory setup for water experiment with motor-cam pump. (Effect of RPM change) ①the DC power supply, ②the dialyzer, ③the motor cam pump, ④the water reservoir

A dialyzer was included in the circuit to provide the same condition as previous experiment. The water from the water reservoir flowed into the pump. The pumped water went through the dialysate side of the dialyzer. The water entered the dialyzer from the bottom inlet and exited at the top outlet. The experiment was conducted on the dialysate on purpose since the dialysate side is harder to pump.

The objective of the experiment is comparing the RPM and flow rate. However, the voltage is controlled to run the pump in certain rpm. Thus, the relationship between the voltage and rpm is checked first.

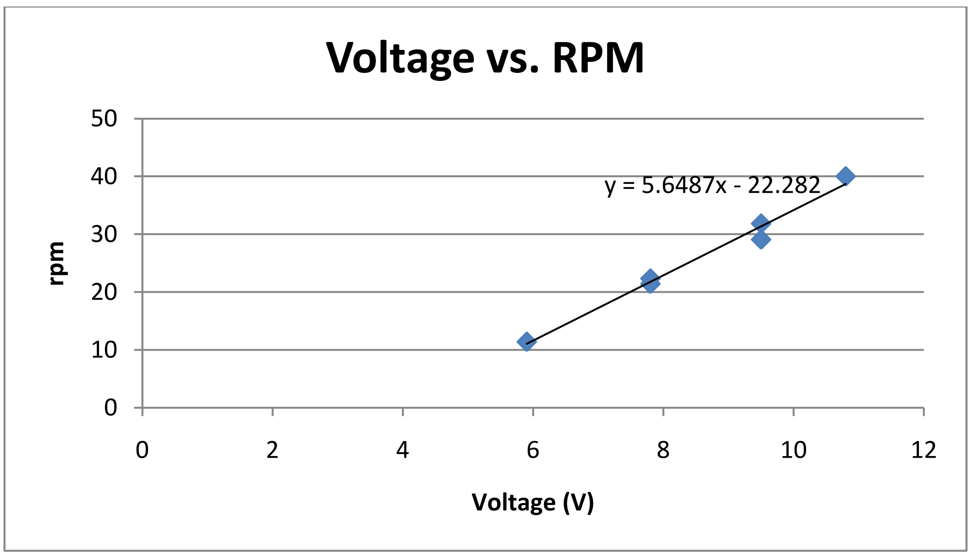


Figure 7-8. [Experiment VI] Voltage vs. RPM

Figure 7-8 shows the relationship between the voltage and rpm. The motor does not run when the voltage is too low due to the lack of torque. Once the motor starts to run, rpm increases linearly with the voltage increase. Since this relation is particular for this motor under a given setting, the equation is implemented in the model to match rpm to the given applied voltage. If a different motor is used, then this experiment should be conducted again and the equation in the model should be changed.

The relationship between rpm and flow rate is shown in Figure 7-9. As was expected, the flow rate increases linearly as the rpm increases. However, it should be noted that the relationship has a small offset (by 0.779) but was ignored in the model for simplicity.

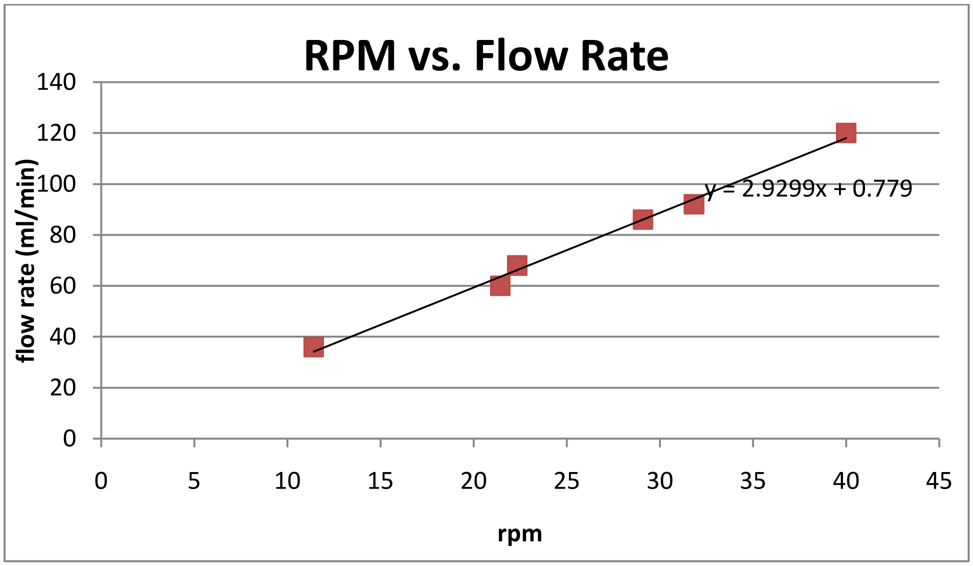


Figure 7-9. [Experiment VI] RPM vs. Flow Rate

7.2 Finding Oval Constant

7.2.1 Motor-Cam Driven Pump Model

The experiment data were used to calibrate the model for the predefined experimental settings . As explained before, oval constant is used as a lumped constant that takes account the head change or back flow loss. For a given pump design, the oval constant changes linearly with the flow rate change. Thus, the oval constant can be determined as follows.

First, model prediction data with the default oval constant 0.785 are evaluated as shown in the second part of Table 7-6. (The Table shows experiment results as well.)

Table 7-6. [Experiment VI] RPM vs. Flow rate

		RPM vs. Flow Rate					
Experiment results	voltage	5.9	7.8	7.8	9.5	9.5	10.8
	rpm	11.4068	22.3380	21.4285	31.8302	29.0838	40
	flow rate (ml/min)	36	68	60	92	86	120
Model prediction (0.785)	rpm	11.05	21.78		31.38		38.7
	Flow rate (ml/min)	46	90		130		160
Model prediction (0.589)	Flow rate (ml/min)	34.5	68		97		120

Then, the values are plotted with experiment data as shown in Figure 7-10.

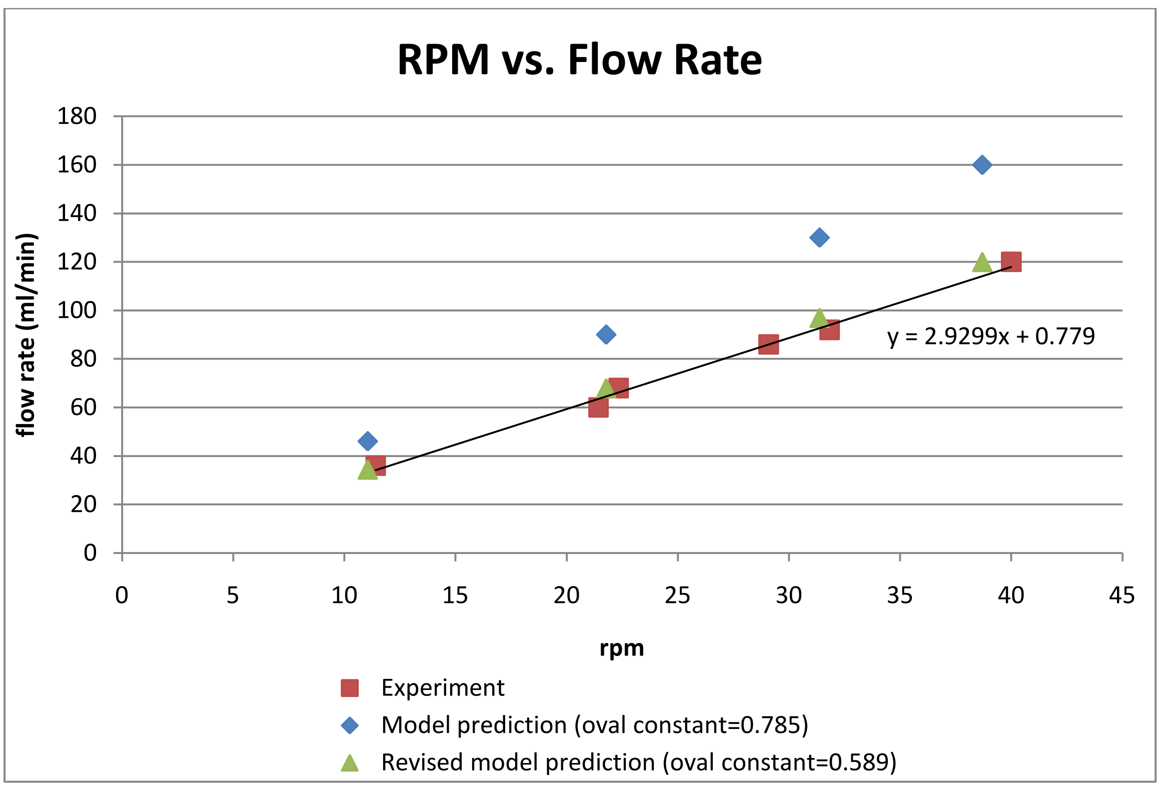


Figure 7-10. Finding oval constant

The figure shows the difference that should be reduced. The next step would be finding a revised oval constant that will reduce the gap between the model prediction and experiment data. The oval constant can be determined from a simple flow rate comparison at a certain voltage. In this case, flow rates at 7.8V were used. (68ml/min:90ml/min=new oval constant:0.785) The figure shows the revised model prediction with the new oval constant 0.589 that matches very well with the experiment data.

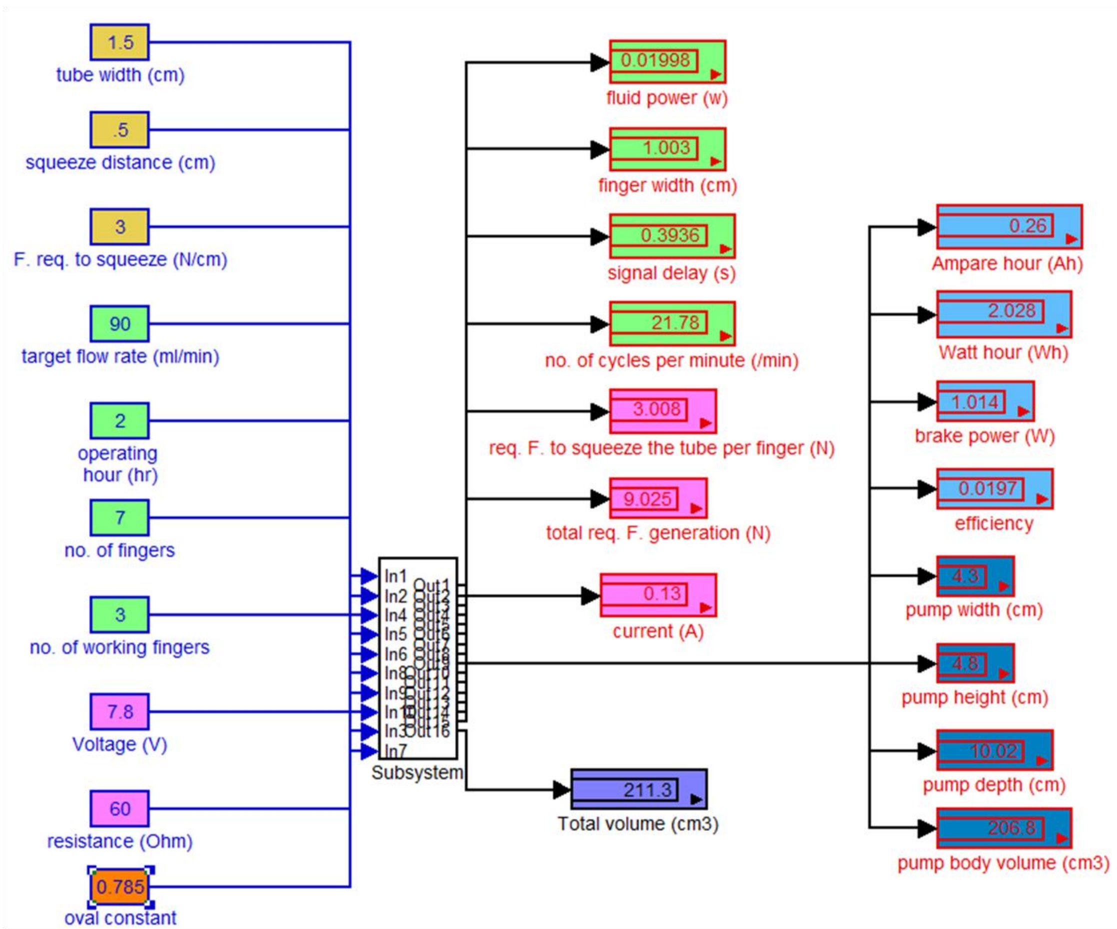


Figure 7-11. Motor-cam driven pump model result with oval constant 0.785

Figure 7-11 and Figure 7-12 shows the pump model results before and after the oval constant revision. Except for the flow rate, fluid power, and efficiency, others are kept the same which means that the model can predict the performance of the pump right. (Efficiency change is due to the fluid power change started from the different flow rate.)

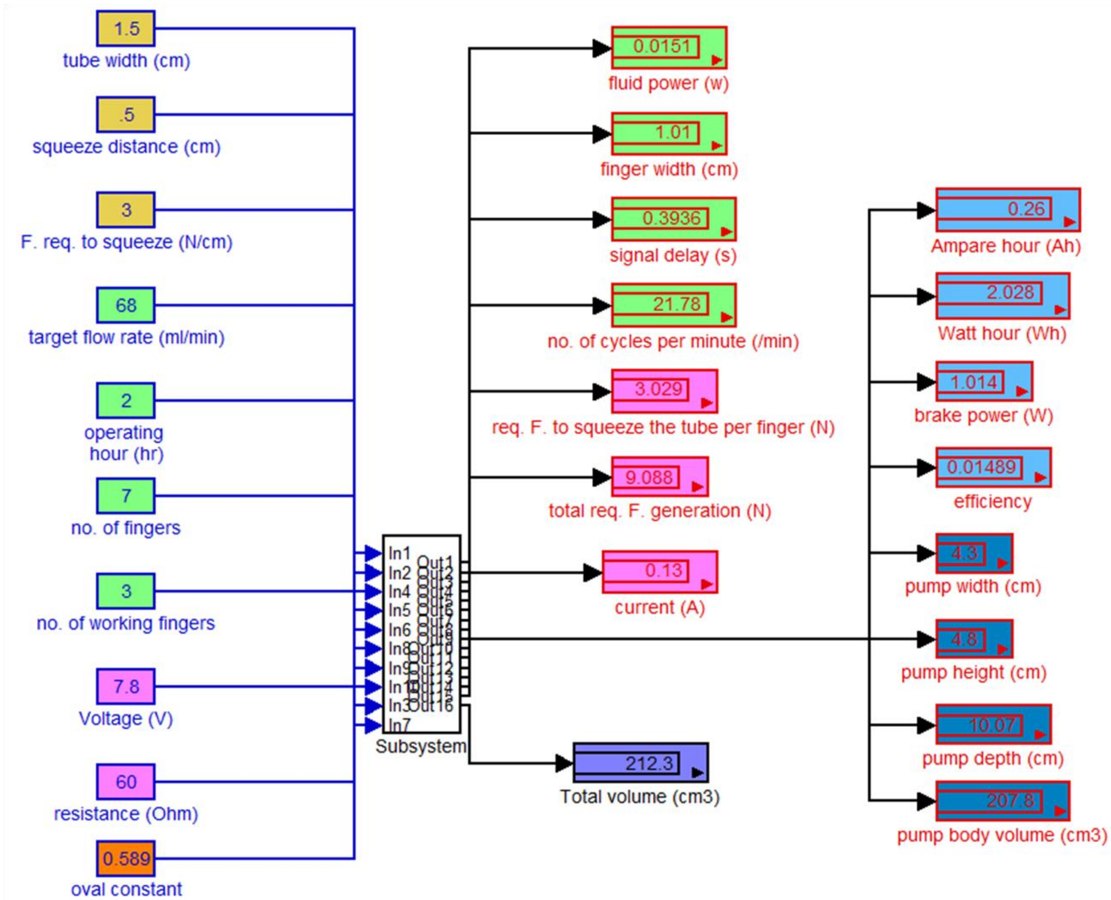


Figure 7-12. Motor-cam driven pump model result with oval constant 0.589

The oval constant 0.589 is the right value for the in vitro experiment setting and will be used to find a better design in the next chapter.

If the pump needs to run in different configuration that would change the flow rate of the pump, the same process can be repeated to find a new oval constant.

7.2.2 Flexinol Driven Pump Design

Flexinol driven pump is tested under the setting shown in Figure 7-13.

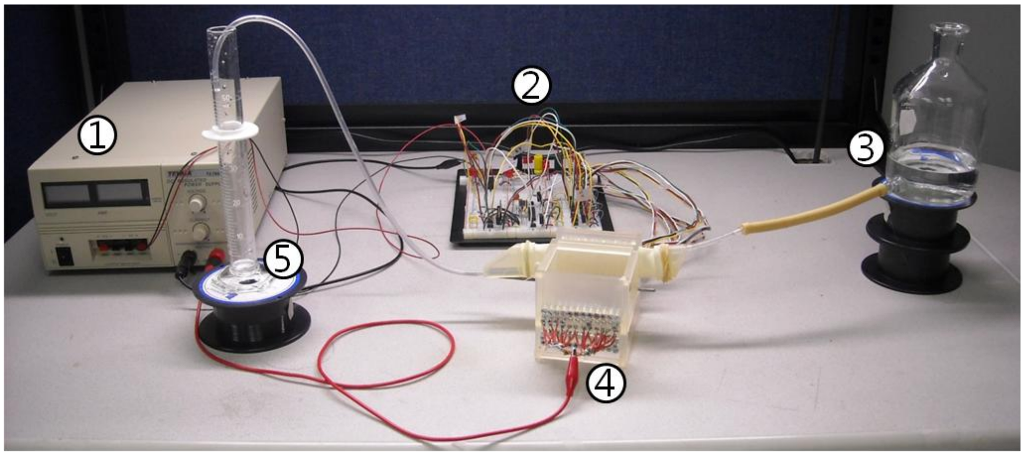


Figure 7-13. The Laboratory Setup for Water Experiment with Flexinol Pump. ①the DC power supply, ②the control circuit, ③the water reservoir, ④the Flexinol pump, ⑤the water disposal

The flow rate was 24ml/min under the pump design parameters shown in Figure 7-14.

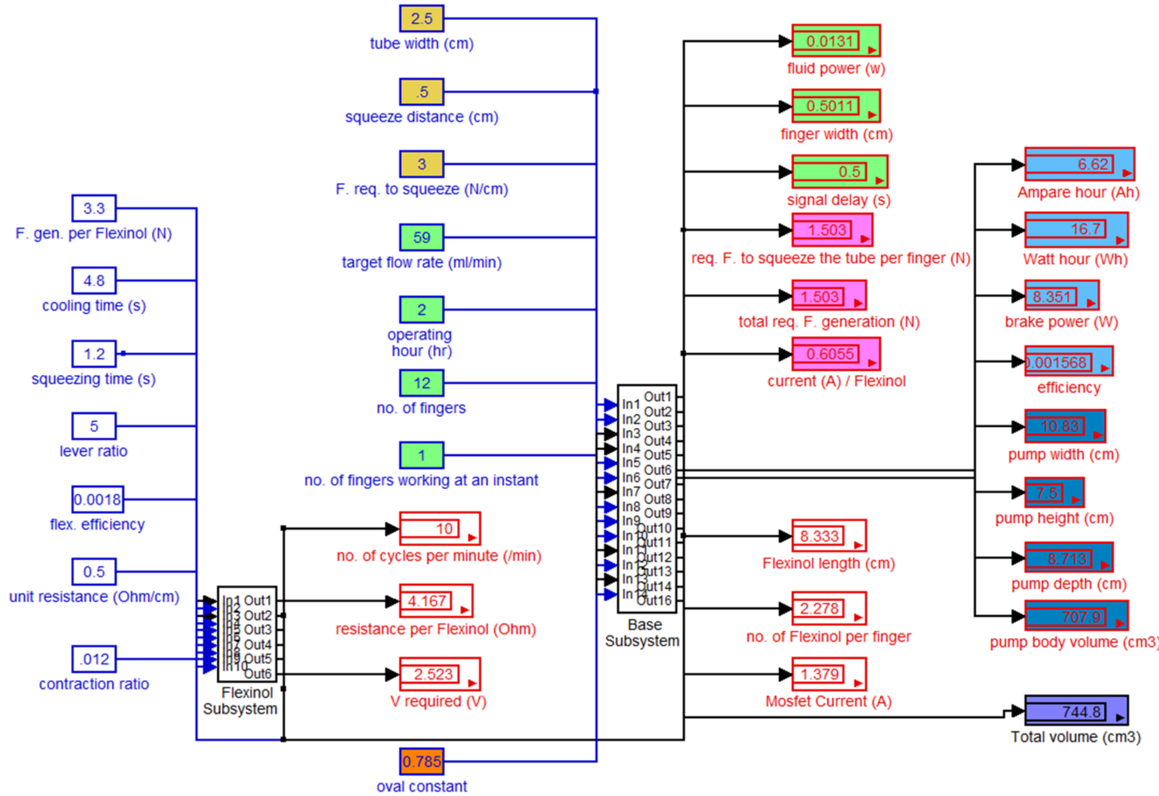


Figure 7-14. Flexinol driven pump model result with oval constant 0.785

It should be noted that the Flexinol efficiency of 0.0018, contraction ratio of 0.012, and cooling time of 4.5 seconds are used. Although according to the data sheet, those were 0.05, 0.03, and 1.3 seconds, the actual performance of Flexinol was not as good as the given information.

As shown in the figure, the default oval constant 0.785 predicts 59ml/min of flow rate. The new oval constant 0.32 is found the same way as before. The revised pump model values are shown in Figure 7-15, and it predicts the experimental flow rate 24ml/min. Since the oval constant is a lumped constant that takes care of different experiment settings such as pump head, character of tube, and back flow, the revised oval constant for this Flexinol driven design is different from the one for motor cam driven design. (Head and the type of tube used were different.)

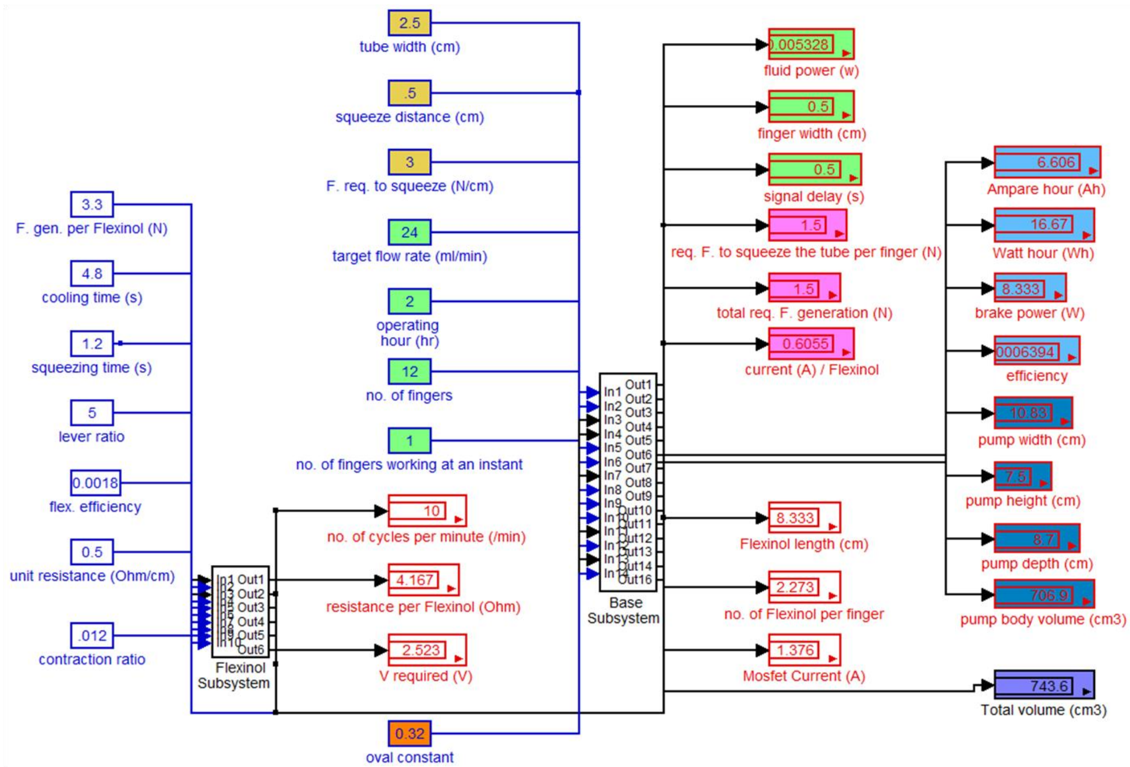


Figure 7-15. Flexinol driven pump model result with oval constant 0.32

7.2.3 Limitation of the Oval Constants

In the previous section, oval constants that represent the predefined experimental setting were found for both designs. The oval constants are proved to match the experiment results with the given pump design. To use the same oval constant for the design space exploration in the next chapter, the pump model should still be able to predict the pump's performance with the same oval constant found in this section for different pump designs such as pumps that have different finger width or different tubing size. In principle, changing other design variables such as tube width or finger width should not affect the oval constant as long as the experiment setting is the same because those design variables' effect on the pump's performance is already included in the pump model which means those are not coupled in theory. In this work, it will be assumed that the same oval constant will work for different designs because the tested pump design is believed to be very close to the smallest pump design possible. Thus, even if other design variables does affect the oval constant, the effect on the oval constant will be negligible in the search range compared to the effect of experiment setting. Thus, the oval constants found in this section will be used to explore the design space in the next chapter.

7.3 Validation of the Pump for the Portable Renal Replacement System

In the previous section, both pump models are validated and oval constants are found that matches the *in vitro* experimental setup. In this section, the motor-cam driven pump is used for *in vitro* experiments to prove that the pump can run the experiment just as currently used pumps for the hemodialysis treatment. Since the difference between the motor-cam driven pump and the Flexinol driven pump is the actuator, not the concept that sequential finger movements pump the fluid, *in vitro* experiment results with the motor-cam driven pump will be enough for the validation of the pump model for the use for a portable renal replacement system.

7.3.1 Laboratory Setup

Porcine blood was obtained during desanguination at a local abattoir, and Citrate and Heparin were added to the blood immediately after collection at a ratio of 100:10:890 (35% Citrate solution: Heparin: blood). Then, the porcine blood was transferred to the laboratory in a thermally insulated container. Dialysate was prepared by mixing concentrates with deionized water at a ratio of 22:38:940 (acid concentrate: bicarbonate concentrate: deionized water).

The blood was pumped into the dialyzer and then came back to the jar. A dialyzer (Gambro Polyflux 6H) was used to perform the dialysis.

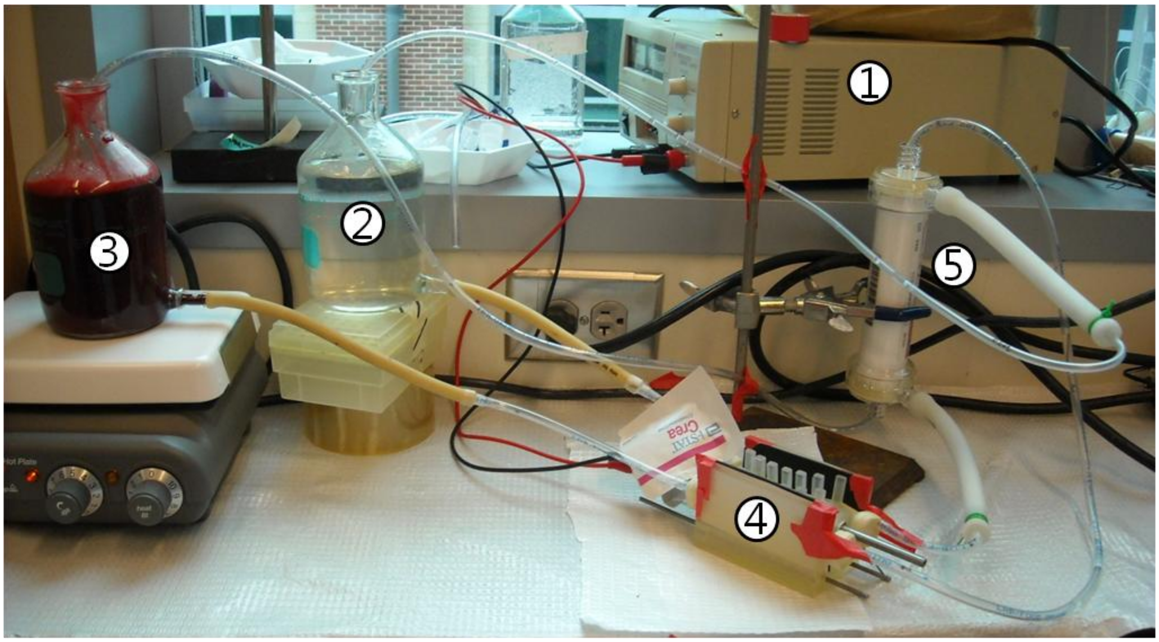


Figure 7-16. The laboratory setup for *in vitro* experiment (recirculation mode) with motor-cam pump. ①the DC power supply, ②the dialysate reservoir, ③the blood reservoir, ④the motor cam pump, ⑤the dialyzer

Figure 7-16 shows the *in vitro* experiment setup conducted using the motor-cam pump. The setting is for the recirculation mode (the dialysate is reused). Recirculation mode requires less dialysate, but lessens the mass transfer rate from the blood due to a smaller concentration gradient.

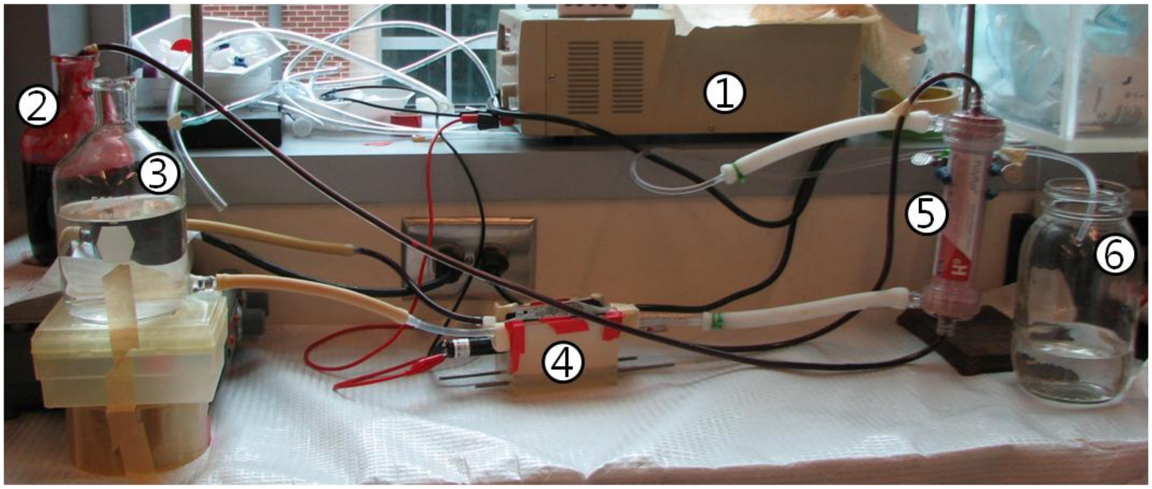


Figure 7-17. The laboratory setup for *in vitro* experiment (single pass mode) with motor-cam Pump. ①the DC power supply, ②the dialysate reservoir, ③the blood reservoir, ④the motor cam pump, ⑤the dialyzer, ⑥the dialysate disposal

Figure 7-17 shows experimental setup for the single pass mode. Used dialysate was flowed out of the dialyzer into the dialysate disposal, and fresh dialysate is frequently poured into the dialysate reservoir to keep the fluid level above the outlet of the jar. As was tested in a previous section, the head difference in the jar does not affect the flow rate, so the flow rate was consistent during the experiment.

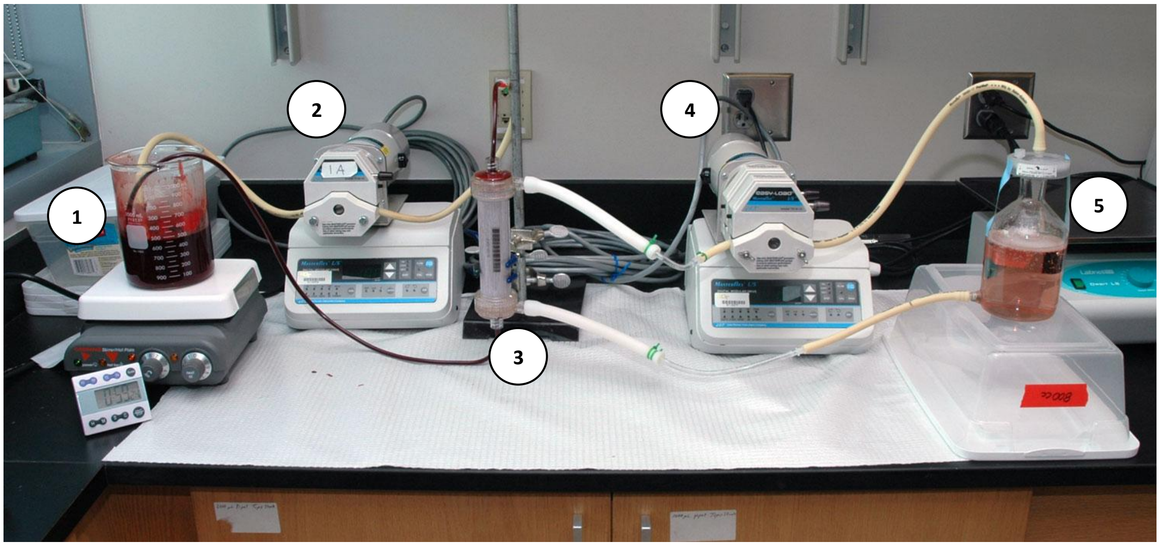


Figure 7-18. The laboratory setup used for experiments. From left to right: the blood reservoir (1), the blood pump (2), the dialyzer (3), the dialysate pump (4), and the dialysate reservoir (5). Olson [6]

Figure 7-18 shows the laboratory setup conducted using the conventional peristaltic pumps. Each pump pumped one fluid, blood and dialysate.

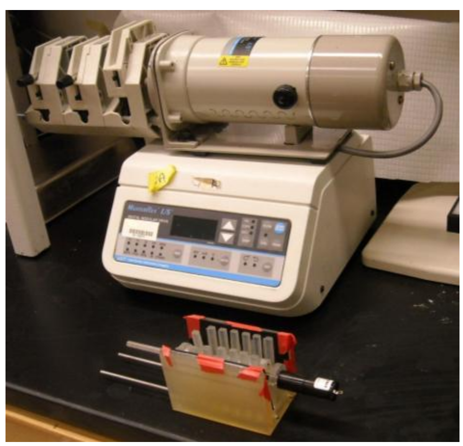


Figure 7-19. The size comparison between the conventional peristaltic pump and the pump designed in this work

Figure 7-19 shows the conventional peristaltic pump and the pump designed in this work. The small pump replaces two of large conventional pumps. Above two figures are shown for the comparison. Although this finger pump uses a DC power supply for now, a small battery can replace the DC power supply. Considering that the small pump is not the minimized pump and even smaller pump design will be proposed in Chapter 8, the size reduction is huge.

7.3.2 Olson's Simulation Package

Olson [5] developed a simulation package that predicts the waste level over time for hemodialysis treatment. The simulation results match well with both *in vitro* experiment data and published patient data. This thesis will compare experiment results with the simulation result from Olson's simulation package to prove the validity of the use of the suggested pump for a renal replacement system.

7.3.3 KoA

The individual solute clearance, K_0, is the measure of clearance of an individual solute. Manufacturers provide K_0A, the variable lumped with the surface area, in the specifications for their dialyzers.

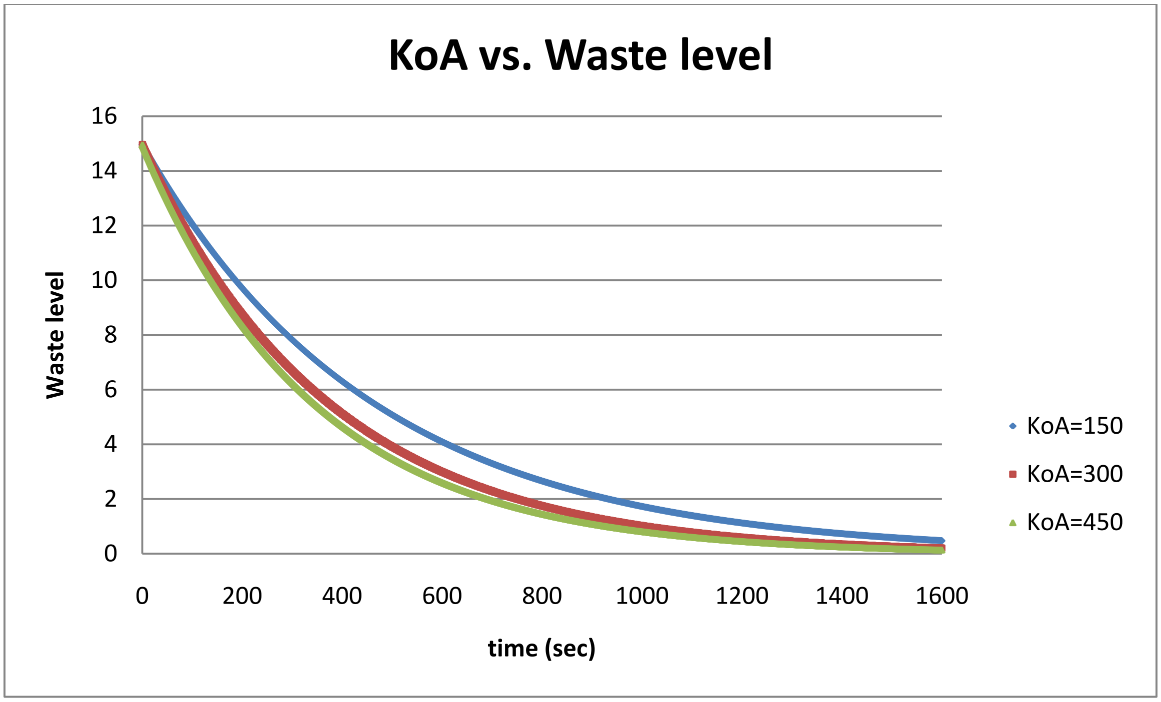


Figure 7-20. Waste level change for different KoA values

Figure 7-20 shows how different KoA values yield different waste removal rate. A dialyzer with a smaller KoA removes waste slower than the one with a larger KoA. Thus, KoA is an important dialyzer characteristic to know for the prediction of the waste level over time. However, KoA is different for each solute for different flow rates, and the manufacturer only provided KoA of Creatinine for the flow rate around 500ml/min, which is much higher than the flow rate of a portable renal replacement system. Thus, KoA value had to be found through an experiment.

With a set of experiment data, the only unknown is KoA. Thus, if more than two data points match the simulation package prediction with a certain KoA value, the value is the one for the solute. The KoA value found for the range of the flow rate lower than 100ml/min will be introduced in the next section, and it will validate the use of the pump for a portable renal replacement system at the same time.Results

7.3.3.1 Single pass mode

A single pass mode experiment was conducted first to find the KoA value. 500ml of blood was used with the blood flow rate of 105ml/min and dialysate flow rate of 100 ml/min. Creatinine level was measured over time with an iSTAT analyzer (Abbott). The experiment data are provided in Table 7-7.

Table 7-7. Creatinine level change over time (single pass mode)

Time (seconds)	Creatinine concentration (mg/dl)
0	11.9
300	6.4
660	3
900	1.9
1140	1.2
1380	0.9

The data points are plotted in Figure 7-21 with the model prediction with KoA, 150.

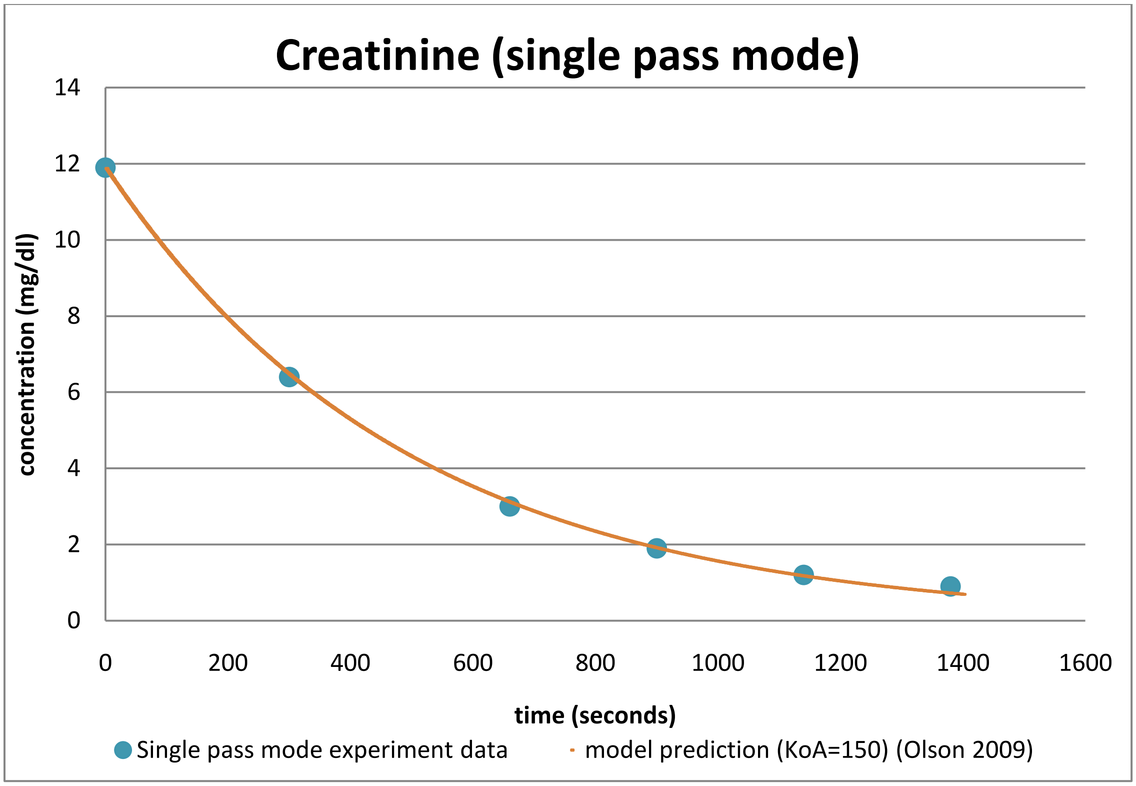


Figure 7-21. Creatinine level change over time (single pass mode)

As can be seen, the experiment data match the prediction value when KoA value is 150. There were a bit of blood leakage in the connection part of the blood tube, but the amount was small enough to say that the experiment is valid. This result found the right value of KoA and validated the pump use for a portable renal replacement system for the single pass mode.

7.3.3.2 Recirculation mode

Another experiment was conducted in the recirculation mode. 450ml of blood ran with the flow rate of 74ml/min, and 550ml of dialysate ran with the flow rate of 75ml/min. Since it is the recirculation mode, the limited amount of dialysate was reused. As before, Creatinine level was measured over time. The experiment data provided in Table 7-8.

Table 7-8. Creatinine level change over time (recirculation mode)

Time (seconds)	Creatinine concentration (mg/dl)
0	13.1
120	11.3
300	8.5
480	7.5
660	6.4
840	6.2
1020	6.7

The data points are plotted in Figure 7-22 with the model prediction with the same KoA, 150.

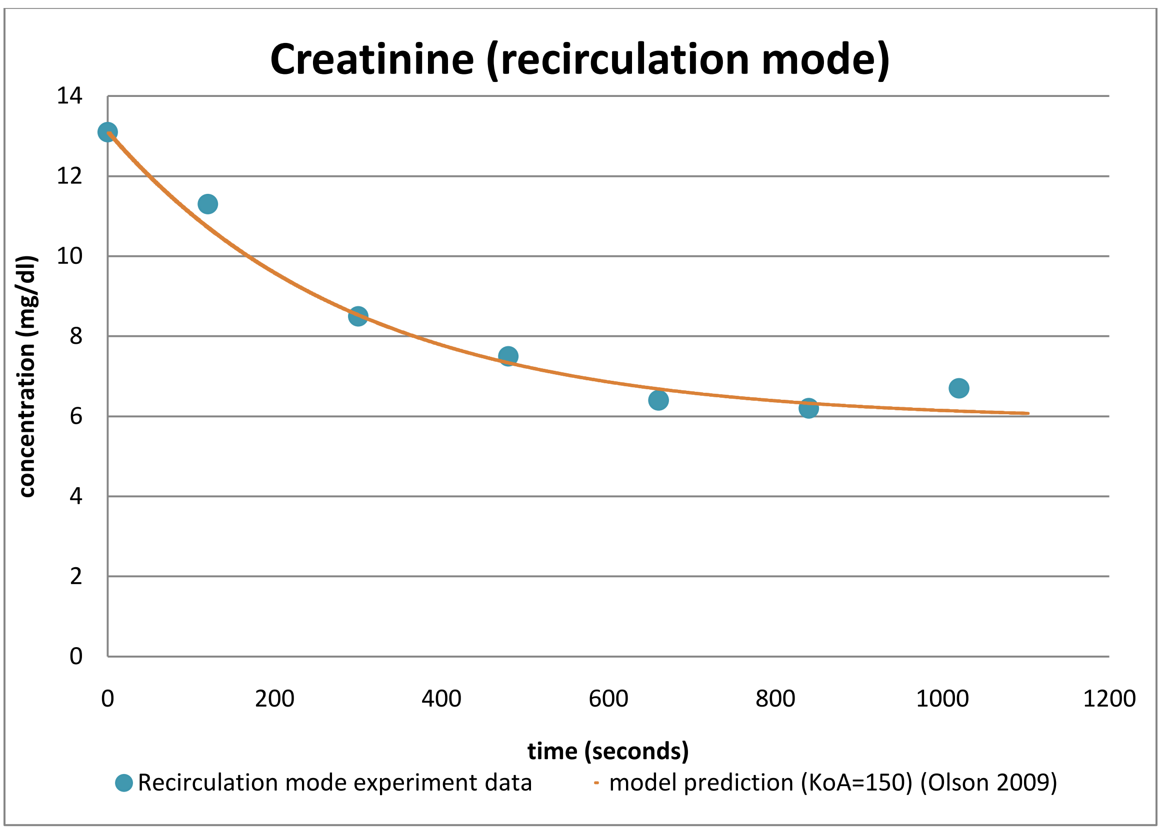


Figure 7-22. Creatinine level change over time (recirculation mode)

Although there were some fluid transfer in the middle of the experiment, the data points matched the prediction values again. This result proves that the pump works well for the recirculation mode of a portable renal replacement system as well.

7.3.4 Conclusion

In this chapter, the performance of the pumps, of which detail designs proposed in Chapter 6, was validated through the *in vitro* experiments. In addition, pump models' ability to predict the performance and specification, such as size and power consumption, of the pump was validated. Oval constants are found for the *in vitro* experiment setting so that both motor-cam driven and Flexinol driven pump models match the experiment setting.

The research question this chapter focused on was:

> *Can an analytical model that embeds relationships between limited number of design parameters (which includes design requirements, design specifications, and characteristics of an actuator) be reliably used to explore the design space under a predefined experimental setting?*

Reliability of pump models were proven with the oval constant found from experiments. Now, the pump models are ready to be used to explore the design space and find the pump design that best fits a portable renal replacement system in the following chapter.

CHAPTER EIGHT

PROPOSED PUMP DESIGN FOR THE PORTABLE RENAL REPLACEMENT SYSTEM

The previous chapter presented both motor-cam and Flexinol driven pumps built and several experiment results that validated the pump models. In the process, the oval constants were found to match the prediction of models to the performance of the pump in the given experiment setting which is close to a renal replacement system. This chapter will use the models with the new oval constants to propose a finalized pump design.

First, the actuator will be finalized by comparing two minimized designs for both actuators. With the finalized actuator, additional design space exploration will be conducted to find out the best design for different objectives other than the size minimization. Finally, one design will be proposed as the best design for a portable renal replacement system.

8.1 Finalizing Actuator for a Portable Renal Replacement System

In this section, design space will be explored to find out the smallest designs for both actuators. Comparing two smallest designs, the actuator will be finalized for the pump design.

8.1.1 Minimized Flexinol Driven Pump Design

The problem of the Flexinol driven pump design is formulated as below.

Given: The Flexinol pump model created in Chapter 5 with the oval constant found in Chapter 7.

Find: The input variable combination for the pump design.

(Tube width, squeeze distance, number of fingers, lever ratio)

Satisfy: 1. Finger width >0.4cm: As explained before, the finger width should be over the critical length to increase efficiency and provide the space for washers in between fingers.

2. Number of Flexinols per finger<6: It is desired to keep the number of Flexinols attached to a finger less than 6 due to the manufacturing and cooling reason.

Objective: Minimize the total volume of the pump: The total volume of the pump includes the pump body volume and the battery volume required to operate the pump for 2 hours. Minimizing the total volume of the pump simplifies the objective by converting the energy consumption information into the size information instead of considering both energy consumption and size of the pump body.

According to the problem formulated above, the design space was explored in the region where the input variables ranged as shown in Table 8-1. The output variables were the all variables included in the pump model.

Table 8-1. Input variables for the design space exploration of the Flexinol driven pump design

inputs	Min	Increments	Max
tube width (cm)	3.5	0.1	5
squeeze distance (cm)	0.2	0.1	2
no. of fingers	7	1	25
Lever ratio	10	1	20

The range for the input values was determined based on both designer's knowledge on the design space from the study shown in Chapter 5 and practical reasons. Maximum tube width was determined as 5cm because the flexible penronse drain tube will not be able to stay in the shape if the tube width gets too large. In addition, the model embedded the relationship between force required to squeeze the tube and tube length, but not width. It was assumed that the tube width will be shorter than 5cm. The minimum value was determined based on the designer's knowledge that the Flexinol design requires smaller contraction distance. Ranges for number of fingers and lever ratio were chosen to be rather large, because the designer did not have enough knowledge.

Increments of tube width and squeeze distance were determined based on the fact that 1mm is small enough change for those variables. Although reducing the increments to a even smaller values could lead to a better search results, practically, tubes with 3.5cm width or 3.55cm width will not provide much of difference. Increment of no. of fingers was 1 because it should be natural numbers. Increment of lever ratio was determined with similar reason as that of tube width. Reducing the increment could lead to a better search result, but it was considered to be a small enough value considering that the lever ratio difference of 1 leads to about 0.15cm difference in the pump width.

After exploring the design space in the input variable range shown above, four designs shown in Table 8-4 were found to have total volume smaller than 630cm^3.

Table 8-2. Final results of the Flexinol driven pump design space exploration

flow rate (ml/min)	100	100	100	**100**
tube width (cm)	5	4.9	5	**5**
squeeze distance (cm)	1.3	1.2	1.2	**1.1**
no. of fingers	12	13	13	**14**
lever ratio	16	16	16	**16**
Voltage (V)	1.587901	1.465755	1.465755	**1.343609**
Oval constant	0.32	0.32	0.32	**0.32**
pump body volume (cm^3)	552.7538	551.4463	549.6292	**547.9614**
pump width (cm)	7.3625	6.95	6.95	**6.5375**
pump depth (cm)	7.507692	8.014626	7.908333	**8.381818**
pump height (cm)	10	9.9	10	**10**
Efficiency	0.001279	0.001253	0.001279	**0.001279**
Watt hour (Wh)	34.72222	35.43084	34.72222	**34.72222**
Ampere hour (Ah)	21.86674	24.17242	23.68897	**25.84251**
F. required per finger (N)	1.201923	1.226452	1.201923	**1.217532**
signal delay (sec)	0.5	0.461538	0.461538	**0.428571**
RPM (/min)	10	10	10	**10**
finger width (cm)	0.400641	0.408817	0.400641	**0.405844**
no. of Flexinols per finger	5.827506	5.946435	5.827506	**5.903188**
total volume (cm^3)	629.3684	629.6245	626.2438	**624.5759**

The last design yielded the smallest total volume and was chosen to be the best design. The pump body is 6.5cm×8.4cm×10cm big, and the total pump volume is 624.5759cm^3 including the battery to run the pump for two hours. Efficiency is 0.128% with brake power of 17.36W.

8.1.2 Minimized Motor-Cam Driven Pump Design

The motor-cam driven design went through the same process as before. The problem is formulated as below.

Given: The pump model created in Chapter 5 with the oval constant found in Chapter 7.

Find: The input variable combination for the pump design.

(Tube width, squeeze distance, number of fingers, voltage)

Satisfy: 1. Finger width >0.4cm: As explained before, the finger width should be over the critical length to increase efficiency and provide the space for the washer in between fingers.

2. Pump depth >7.5cm, Pump width >4.3cm, Pump height >4.3cm: Additional restrictions to be satisfied were the size of the pump. The pump design includes a motor inside, so enough space should be left to position a motor. It requires a pump to be larger than 4.3cm wide, 4.3cm high, and 7.5cm deep. Keeping the depth of the pump longer than a certain value is important not only for positioning a motor inside, but for having a long enough tube to reduce the loss coming from the first and last finger stroke (explained in Chapter 4).

Objective: 1. Minimize the total volume of the pump

2. Minimize the required force generation: Although it is the goal for the next step after achieving objective 1, it is still an important goal since the volume of the pump has possibility to be even smaller with a smaller required force generation of the motor.

The reason for the two objectives for the motor-cam driven design is due to the limit of the motor-cam driven pump model that did not embed the relationship between the required force generation and size of the actuator.

According to the problem formulated above, the design space was explored in the region where the input variables ranged as shown in Table 8-3.

Table 8-3. Input variables for the design space exploration of the motor-cam driven design

inputs	Min	Increments	Max
tube width (cm)	0.5	0.1	5
squeeze distance (cm)	0.2	0.1	2
number of fingers	7	1	13
Voltage (V)	9	1.5	12

Maximum tube width was determined as 5cm due to the same reason as before. The maximum squeeze distance was chosen in the similar reason as tube width. If squeeze distance is too long, it will be hard for the flexible tube to stay in the position. Number of fingers was chosen from the fact that at least 7 fingers are required to provide enough fluid filling time after each compression as explained in Chapter 4. In addition, designer's knowledge that too many fingers yield too short finger width decided the maximum value. Minimum voltage value was chosen from the designer's knowledge that too low voltage cannot provide enough rpm, and the maximum value was determined from the fact that the motor's nominal voltage is 12V. Increment of the voltage was determined based on the commercially available batterie's voltage 1.5V.

After exploring the design space in the input variable range shown above, five designs shown in Table 8-4 were found to have the minimum total volume.

Table 8-4. Final results of the motor –cam driven pump design space exploration

flow rate (ml/min)	100	100	100	100	**100**
tube width (cm)	1.3	1.3	1.3	1.3	**1.3**
squeeze distance (cm)	0.6	0.6	0.6	0.6	**0.6**
no. of fingers	7	8	9	10	**11**
Voltage (V)	12	12	12	12	**12**
Oval constant	0.589	0.589	0.589	0.589	**0.589**
pump body volume (cm^3)	161.1207	161.1207	161.1207	161.1207	**161.1207**
pump width (cm)	4.5	4.5	4.5	4.5	**4.5**
pump depth (cm)	7.78361	7.78361	7.78361	7.78361	**7.78361**
pump height (cm)	4.6	4.6	4.6	4.6	**4.6**
Efficiency	0.00925	0.00925	0.00925	0.00925	**0.00925**
Watt hour (Wh)	4.8	4.8	4.8	4.8	**4.8**
Ampere hour (Ah)	0.4	0.4	0.4	0.4	**0.4**
Current (A)	0.2	0.2	0.2	0.2	**0.2**
total F required (N)	6.150356	5.381562	4.78361	4.305249	**3.913863**
RPM (/min)	45.5024	45.5024	45.5024	45.5024	**45.5024**
F. required per finger (N)	2.050119	1.793854	1.594537	1.435083	**1.304621**
signal delay (sec)	0.188373	0.164826	0.146512	0.131861	**0.119874**
finger width (cm)	0.683373	0.597951	0.531512	0.478361	**0.434874**
total volume (cm^3)	171.7119	171.7119	171.7119	171.7119	**171.7119**

Since all five designs satisfy the objective 1, minimize the total volume of the pump, the last design that requires the minimum total force was chosen to be the best design. The pump body is 4.5cm×7.8cm×4.6cm, and the total pump volume is 171.7119cm^3 including the battery to run the pump for two hours. Efficiency is 0.925% with brake power of 2.4W.

8.1.3 Actuator for a Portable Renal Replacement System

In the previous sections, smallest designs are suggested for each actuator by exploring design spaces using the analytical pump model. Table 8-5 summarizes the specification of each suggested designs.

Table 8-5. Best designs for each actuator

	Flexinol	Motor-cam
total pump volume (cm^3)	625	172
efficiency (%)	0.128	0.925
brake power (W)	17.36	2.4

It is clear that a motor-cam driven design is much better than a Flexinol driven pump. It pumps the same amount of flow while consuming 7.23 times less energy with 3.6 times smaller design. Thus, motor is a better-suited choice than Flexinol for a renal replacement system.

8.2 Finalizing the Pump Design for a Portable Renal Replacement System

In this work, the ojective was to minimize the size of the whole system. Thus, motor is chosen as the final actuator due to its ability to yield a smaller pump design than Flexinol. However, a designer might prefer focusing on other objectives such as minimizing the mass or maximizing the efficiency. In this section, the best designs for those different objectives were searched with the motor-cam driven pump design.

8.2.1 Minimizing the Mass of the Motor-Cam Driven Pump

The problem is formulated the same way as before (section 8.1.2) except that the first objective changed from minimizing the size to minimizing the mass.

However, to mimimize the mass of the system, the density information for the pump body and battery is required. Battery density information is available from commercial battery search in section 3.2.5.4. (The mass of the motor, 60 grams, is not considered assuming that the kind of motor used for the design is anyway the same. Because the mass will be compared for each different design to find the lighter design, the absolute value is not important during the searching process.) The pump body has smaller density than the battery since it includes lots of empty space in it. From section 3.2.5.4, the density of the battery is found to be 2.3 g/cm^3. First, 1.2 g/cm^3 is used for the density of the pump body, which is the value expected to be close to the final density for the commercialization.

Table 8-6. Final results of the motor-cam driven pump design space exploration for the minimization of mass with pump body density 1.2g/cm^3

flow rate (ml/min)	100	100	100	100	**100**
tube width (cm)	1.3	1.3	1.3	1.3	**1.3**
squeeze distance (cm)	0.6	0.6	0.6	0.6	**0.6**
no. of fingers	7	8	9	10	**11**
Voltage (V)	12	12	12	12	**12**
Oval constant	0.589	0.589	0.589	0.589	**0.589**
pump body volume (cm^3)	161.1207	161.1207	161.1207	161.1207	**161.1207**
pump width (cm)	4.5	4.5	4.5	4.5	**4.5**
pump depth (cm)	7.78361	7.78361	7.78361	7.78361	**7.78361**
pump height (cm)	4.6	4.6	4.6	4.6	**4.6**
Efficiency	0.00925	0.00925	0.00925	0.00925	**0.00925**
Watt hour (Wh)	4.8	4.8	4.8	4.8	**4.8**
Ampere hour (Ah)	0.4	0.4	0.4	0.4	**0.4**
Current (A)	0.2	0.2	0.2	0.2	**0.2**
total F required (N)	6.150356	5.381562	4.78361	4.305249	**3.913863**
RPM (/min)	45.5024	45.5024	45.5024	45.5024	**45.5024**
F. required per finger (N)	2.050119	1.793854	1.594537	1.435083	**1.304621**
signal delay (sec)	0.188373	0.164826	0.146512	0.131861	**0.119874**
finger width (cm)	0.683373	0.597951	0.531512	0.478361	**0.434874**
total volume (cm^3)	171.7119	171.7119	171.7119	171.7119	**171.7119**
total mass (grams)	217.7046	217.7046	217.7046	217.7046	**217.7046**

Table 8-6 shows the mass minimization result with the pump body density 1.2g/cm^3. The result came out to be the same as Table 8-4, which was the result for the minimization of the size. It is because the size of the pump body is anyway much bigger than the battery.

Testing different pump body densities, it was found that the best pump for the minimization of the size and mass is the same as long as the pump body density is bigger than 0.6g/cm^3.

Table 8-7. Final results of the motor-cam driven pump design space exploration for the minimization of mass with pump body density 0.6g/cm^3

flow rate (ml/min)	100	100	100	100	**100**
tube width (cm)	1.4	1.4	1.4	1.4	**1.4**
squeeze distance (cm)	0.7	0.7	0.7	0.7	**0.7**
no. of fingers	7	8	9	10	**11**
Voltage (V)	10.5	10.5	10.5	10.5	**10.5**
Oval constant	0.589	0.589	0.589	0.589	**0.589**
pump body volume (cm^3)	169.6195	169.6195	169.6195	169.6195	**169.6195**
pump width (cm)	4.7	4.7	4.7	4.7	**4.7**
pump depth (cm)	7.678564	7.678564	7.678564	7.678564	**7.678564**
pump height (cm)	4.7	4.7	4.7	4.7	**4.7**
Efficiency	0.012082	0.012082	0.012082	0.012082	**0.012082**
Watt hour (Wh)	3.675	3.675	3.675	3.675	**3.675**
Ampere hour (Ah)	0.35	0.35	0.35	0.35	**0.35**
Current (A)	0.175	0.175	0.175	0.175	**0.175**
total F required (N)	6.015296	5.263384	4.678564	4.210707	**3.827916**
RPM (/min)	37.02935	37.02935	37.02935	37.02935	**37.02935**
F. required per finger (N)	2.005099	1.754461	1.559521	1.403569	**1.275972**
signal delay (sec)	0.231477	0.202542	0.180037	0.162034	**0.147303**
finger width (cm)	0.668366	0.58482	0.51984	0.467856	**0.425324**
total volume (cm^3)	177.7284	177.7284	177.7284	177.7284	**177.7284**
total mass (grams)	120.4221	120.4221	120.4221	120.4221	**120.4221**

As shown in Table 8-7, the best design changes when the pump body density is 0.6g/cm^3. The design configurations slightly changed with slight increase in the total pump size compared to the best design for the size minimization. However, it is hard to expect the pump body density to be such small value. These results prove that the best designs for the minimization of the size or mass lies in a small range of the design space. Thus, whether one focuses on the mass or size, the final designs are the same or very similar to each other.

8.2.2 Maximizing Efficiency of the Motor-Cam Driven Pump

The problem is formulated the same way as before except that the objectives are replaced to 1. Maximize efficiency, 2. Minimize the total pump volume, and 3. Minimize the required force generation.

Table 8-8. Final results of the motor-cam driven pump design space exploration for the maximization of efficiency

flow rate (ml/min)	100	100	100	100	**100**
tube width (cm)	2.3	2.3	2.3	2.3	**2.3**
squeeze distance (cm)	1.4	1.4	1.4	1.4	**1.4**
no. of fingers	7	8	9	10	**11**
Voltage (V)	6	6	6	6	**6**
Oval constant	0.589	0.589	0.589	0.589	**0.589**
pump body volume (cm^3)	257.614	257.614	257.614	257.614	**257.614**
pump width (cm)	6.1	6.1	6.1	6.1	**6.1**
pump depth (cm)	7.541393	7.541393	7.541393	7.541393	**7.541393**
pump height (cm)	5.6	5.6	5.6	5.6	**5.6**
Efficiency	0.037	0.037	0.037	0.037	**0.037**
Watt hour (Wh)	1.2	1.2	1.2	1.2	**1.2**
Ampere hour (Ah)	0.2	0.2	0.2	0.2	**0.2**
Current (A)	0.1	0.1	0.1	0.1	**0.1**
total F required (N)	5.838934	5.109068	4.541393	4.087254	**3.715686**
RPM (/min)	11.6102	11.6102	11.6102	11.6102	**11.6102**
F. required per finger (N)	1.946311	1.703023	1.513798	1.362418	**1.238562**
signal delay (sec)	0.738267	0.645984	0.574208	0.516787	**0.469806**
finger width (cm)	0.64877	0.567674	0.504599	0.454139	**0.412854**
total volume (cm^3)	260.2618	260.2618	260.2618	260.2618	**260.2618**

Table 8-8 shows that the results are quite different from the one for mass or size minimization. Efficiency increases when the motor runs in lower rpm, and with decreased voltage, 6V instead of 12V, efficiency increased from 0.925% to 3.7%. The results prove that in case the power consumption is very important, one should sacrifice the total volume of the pump (the total pump volume increased from $172cm^3$ to $260cm^3$).

8.2.3 Proposed Pump Design for a Portable Renal Replacement System

With the pump body density of $1.2g/cm^3$, battery density of $2.3g/cm^3$, and motor mass 60grams, the best designs for different objectives are found as shown in Table 8-9.

Table 8-9. Best pump designs for different objectives

	Min. size	Min. mass	Max. efficiency
total pump volume (cm^3)	**172**		260
total pump mass (grams)	**278**		375
efficiency (%)	**0.925**		3.7
brake power (W)	**2.4**		0.6

One can choose either design or find another best design with different weights on mass and efficiency depending on the need. In this work, the design with minimized size is proposed as the final design since the brake power is low enough.

Applying 10% safety net, the pump will be smaller than 4.5cm×8.6cm×4.6cm with the total volume of $188cm^3$ and power consume of less than 3W. Compared to the smallest pump of a portable renal replacement system in the literature [7] that is 317 cm^3 large and consumes less than 10W of power, the proposed pump design achieved reduction in size by 40% and savings in energy consumption by 65% with the removal of valves.

$$\text{reduction in size} = \frac{(317 - 188)}{317} \times 100 = 40.694\ (\%) \tag{8-1}$$

$$\text{reduction in power consumption} = \frac{(9 - 3)}{9} \times 100 = 66.67\ (\%) \tag{8-2}$$

This simple and reliable design substantially will enable development of a portable renal replacement system.

8.3 Conclusion

In this chapter, the design space is explored and the best pumps were found using the analytical models with the oval constant found for the given experiment setting. These results are valid as long as the predefined experiment setting yields the same oval constant used in this chapter. If the experiment setting changes, then the best pump can be found by following the same steps after finding out the new oval constant with experiments.

This chapter answered to both research questions.

> **Research Question 1:**
>
> *Can a pump that is more compact, lighter, and more energy efficient than currently existing pumps for the portable renal replacement system be designed with lower manufacturing cost, less noise, and less possibility of clogging?*

Adopting a different working principle (finger pump) with a conventional actuator, motor, allowed a better pump with the same level of noise compared to currently existing pumps. Adopting a different working principle with a different actuator, Flexinol, allowed a pump that involves lower manufacturing cost, less noise, and less possibility of clogging. However, it failed to be as compact, light, and energy efficient as the motor-cam driven design.

> **Research Question 2:**
>
> *Can an analytical model that embeds relationships between limited number of design parameters (which includes design requirements, design specifications, and*

characteristics of an actuator) be reliably used to explore the design space under a predefined experimental setting?

This chapter showed that the analytical pump models that embeds relationships between limited number of design parameters are useful and good enough for finding the pump design that best fits the objective with the oval constant for the given setting.

CHAPTER NINE

CLOSURE AND CONTRIBUTIONS

The broad goal of this thesis was to design a novel pump for a portable renal replacement system. Chapter 3 went through the conceptual design and embodiment design phase of the Pahl & Beitz design method that led to the finger pump concept. Chapter 4 provided in-depth understanding of the finger pump. Chapter 5 created the analytical pump model that was used in Chapter 8 to propose a pump design for a portable renal replacement system. Chapter 6 provided the detail design of the pump and Chapter 7 validated the design and the pump model.

In Chapter 1, two research questions were posed for this thesis with a hypothesis with each question. In the first section of this chapter, each hypothesis will be addressed in the context of the work completed in this thesis. The following sections will address the contribution this research has made and specify the scope and limitation of this research. The last section will provide the future work.

9.1 Answering the Research Questions

This research was divided into two research questions, which are now tested.

9.1.1 Research Question One

The first research question was addressed in Chapter 3, 6, and 8.

Research Question 1:

Can a pump that is more compact, lighter, and more energy efficient thacurrently existing pumps for the portable renal replacement system be *designed with lower manufacturing cost, less noise, and less possibility of clogging?*

Hypothesis 1:

By adopting a different working principle and a different actuator, we can design a pump that is more compact, lighter, more energy efficient than currently existing pumps for the portable renal replacement system with lower manufacturing cost, less noise, and less possibility of clogging.

In Chapther 3, a finger pump was found to be the most appropriate working principle to adopt for a portable renal replacement system since it can remove check valves and friction loss while adopting compliant tube. This led to a smaller actuator, which further made the pump more compact, lighter, more energy efficient with lower manufacturing cost and possibility of clogging compared to currently existing pumps. Chapter 3 also concluded a motor and Flexinol as possible actuators. Based on the layout suggested in Chapter 3, detail designs of pumps for both actuators were provided in Chapter 6. However, after building the Flexinol driven pump, running the experiment to find the oval constant, and finding the best design in Chapter 8, the Flexinol driven pump was found to be not as good as the motor-cam driven design. The Flexinol driven design might be a good choice once the performance of the Flexinol improves and matches the performance given in the data sheet. The main reason for the large size and energy consumption of the Flexinol driven design is the low efficiency of Flexinol, less than 0.2%. Once the efficiency of the Flexinol improves so that small enough Flexinol driven pump design is achievable, Flexinol driven pump design could be a better choice than the motor-cam driven pump design since there is no noise at all compared to the motor-cam driven design.

In conclusion, adopting both a different working principle and a different actuator were attempted to design a better pump for a portable renal replacement system. By adopting a different working principle, we were able to propose a novel pump design . However, it was achieved using a motor, which is a conventional actuator, and the pump is still noisy. Adopting a different actuator, Flexinol, did not provide a better design, but it has the potential to be a better design with no noise as the performance of the Flexinol improves.

9.1.2 Research Question Two

The second research question was addressed in Chapter 5, 7, and 8.

Research Question 2:

Can an analytical model that embeds relationship between limited number of design parameters (which includes design requirements, design specifications, and characteristics of an actuator) be reliably used to explore the design space under a predefined experimental setting?

Hypothesis 2:

An analytical model that embeds relationship between limited number of design parameters can be reliably used to explore the design space by introducing a lumped constant that accounts for the predefined experimental setting.

Following the basic relationships that govern a finger pump specifications and relating those to design requirements and characteristics of an actuator in Simulink, MATLAB, an analytical model was created in Chapter 5. Using the models, the relationship between input and output variables is investigated. A lumped constant (oval constant) was included as a parameter of the model to consider the experimental setting. The lumped constant represents effects coming from other than those parameters included in the model. In Chapter 7, the model's ability to predict the pump's performance and specifications was validated by conducting experiments with the pumps. It proved that an oval constant can be calibrated for an experimental setting, and with the oval constant, the analytical model can provide reliable prediction. Then, the model was used to explore the design space in Chapter 8 proposing the final design.

9.2 Contributions

The overall goal of this thesis was to design a novel pump for a portable renal replacement system. In that scope, several contributions were made.

9.2.1 Systematic Analysis of Positive Displacement Pumps

Working principles of positive displacement (PD) pumps are analyzed in Chapter 3 according to the Pahl & Beitz method. PD pumps are differentiated according to the applied force direction (transient and normal), which made it possible to understand where the energy and power consumption comes from and to reduce those factors. PD pump working principle analysis will be a useful reference for designing any other PD pumps.

9.2.2 An Analytical Finger Pump Model

This thesis created an analytical finger pump model that embeds relationships between design parameters that affect the specification of the pump. In the process of creating the model, the relationships of finger pump design parameters were investigated and understood. Although the model should be matched to a predefined experimental setting, once it is set to the setting by finding the oval constant, the model can be reliably used for predicting the performance of the pump. The pump model could be used for designing any other finger pumps that have different requirement such as flow rate. In addition, experiments proved that a pump can be analytically modeled to be reliable enough for the design purpose.

9.2.3 Adoptation of a Flexible Tube

No existing PD pumps use a flexible tube since the tube restoring force is responsible for making a vacuumed space after compression that forces the fluid to fill the space so that it can be pushed in one direction. However, this pump design adopted the flexible tube to reduce the force requirement for an actuator and enabled to design a smaller pump. Although using the flexible tube limits the ability of the pump for suction, as long as the purpose of the pump does not require suction, using flexible tube is a great step towards the miniaturization of a pump.

9.2.4 Proof of Concept for a Finger Pump

Although the finger pump existed for long time, most of clinically used pumps that require keeping the blood circuit clean uses peristaltic pump. This thesis proved that a finger pump can be a substitute for those peristaltic pumps.

9.2.5 Design of a Pump for a Portable Renal Replacement System

The most important goal of this thesis was designing a novel pump. This thesis proposed a pump that can pump two different fluids at a time with design that is smaller than $188cm^3$ and consumes less than 3W. It is reduction in size by 40% and savings in energy consumption by 65% compared to the best pump in the literature. In addition, check valves inside the tube were removed, and it will lead to lower manufacturing cost and less possibility of clogging. This miniaturized pump will substantially enable development of a portable renal replacement system. The portable system will improve ESRD patients' quality of life by allowing them to get the treatment home without visiting the clinic. In addition, it will be able to provide more frequent treatment that can lower the waste level and improve patients' health.

9.3 Scope and limitations of this research

One major limitation of the proposed pump design is lacking ability for suction. Suction is not required for a portable renal replacement system, so this pump is good for this application. However, if suction is required for other application, a stiff tube that can restore the shape after compression on its own should be used. Using a stiff tube will increase the force requirement for the motor thus increasing the size of the motor. As a result, the same working principle can be used for designing a pump that requires suction, but the size and energy consumption will increase.

Another limitation of the proposed pump design is the target flow rate. The pump will be able to generate up to 100ml/min for the given input voltage 12V. Lower flow rates can easily be generated by applying lower voltage or using smaller tube, but if a higher flow rate is required, the pump will either be bigger or consume more energy due to higher required input voltage.

A limitation of the analytical pump model is that the model can work only when the designer already knows the operating condition of the pump. The oval constant for each different operating condition should be found with experiments to model the pump.

9.4 Future work

The following directions for future work have been identified for realizing a portable renal replacement system.

9.4.1 Two Shaft Design

In Chapter 6, a two shafts design that puts the motor inside the pump was proposed. However, it was not built, yet. The two shafts design will have to follow before this pump goes into the market.

9.4.2 Tube Insertion Mechanism

The proposed pump design lacks a tube insertion mechanism which was mentioned in section 4.1 with Figure 4-4. Because at least one finger is in the position of occluding the tube, it is not easy to insert the tube. This mechanism can facilitate the tube insertion process for patients.

9.4.3 Pressure Sensor and Leakage Detector

In the middle of treatment, the pressure could increase due to a clogging or leaks may start due to a problem in the flow circuit connection. In that case, the pump should automatically stop operating. Thus, a pressure sensor and leakage detector should be added in the flow circuit and connected to the motor control unit to enable this feature.

9.4.4 Flow Profile Analysis

The finger pump generates a pulsatile flow. Thus, the flow profile should be analyzed to understand how the flow rate changes during a compression cycle. It was not a necessary information to know for designing the pump as long as the overall flow rate can be predicted, it will be an important characteristic of the finger pump to include in the data sheet. In addition, the flow profile analysis could lead to understanding of how the signal delay between each fingers affects the flow profile. Once it is understood, one can control the signal delay to generate the desired flow profile by changing the cam design.

9.4.5 Shear Stress Analysis

This thesis proved that the finger pump can be used for renal replacement system by providing *in vitro* experiment data with enough of flow rates and Creatinine level change over time. However, the shear stress analysis is still required to make sure that the finger pump does not involve blood cell damage over limit.

9.4.6 Tube Design

This thesis used penrose drain tube since it is the most flexible tube available in the market. However, there are limited size of tubes offered, and the proposed design requires a tube size that is not available. Thus, a flexible tube that has the required size for this pump should follow.

9.4.7 Prototype of a Portable Renal Replacement System

The pump was designed for a portable renal replacement system. Thus, the prototype should be made and run with a battery, not a DC power supply.

APPENDIX A

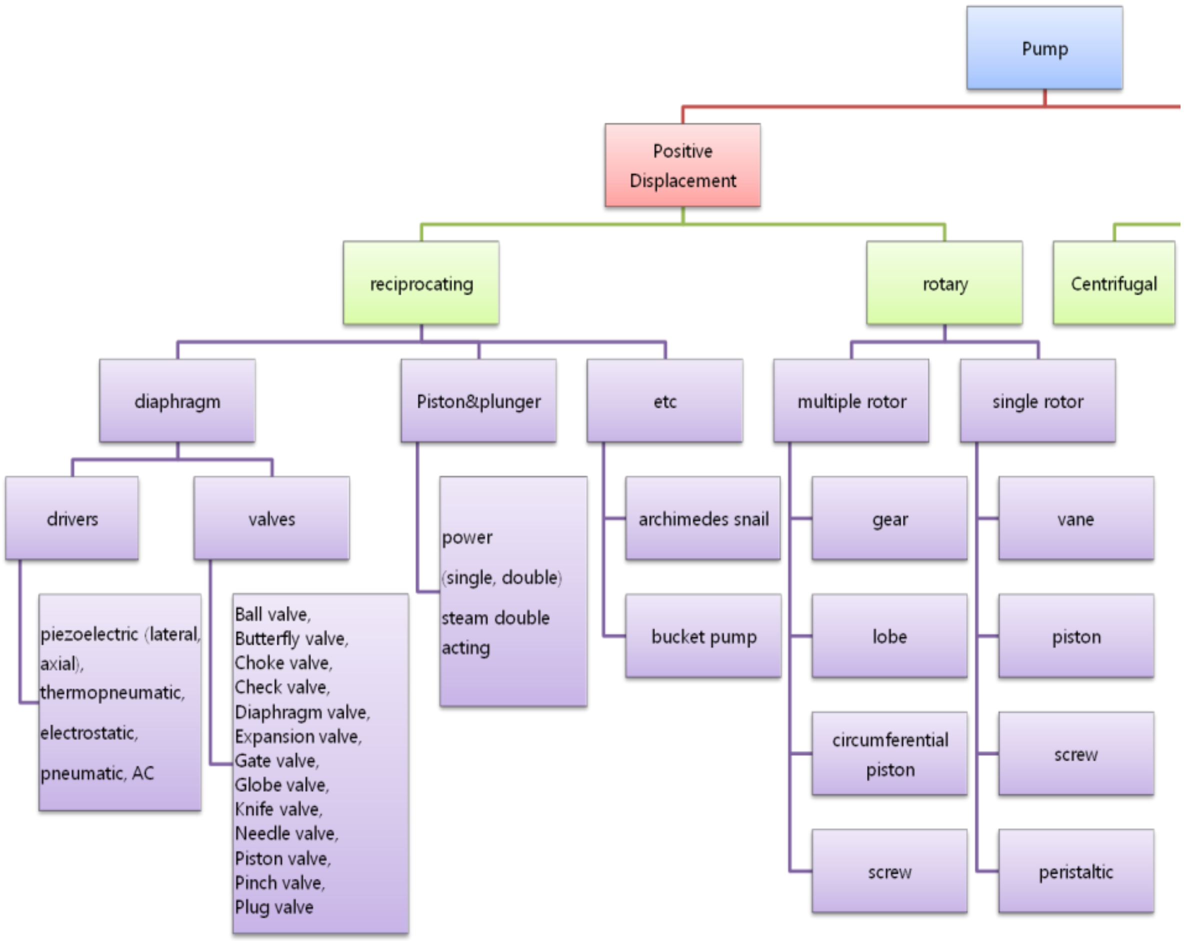
GLOSSARY OF TERMS

brake power	The power required to operate a pump. Different from *fluid power*, which is the power associated with the moving fluid.
centrifugal pump	A pump that relies on an impeller inside the fluid to provide a pressure rise.
check valves	a mechanical device, a valve, which normally allows fluid to flow through it in only one direction.
citrate	A common blood anticoagulant.
dialysate	The solution that runs on the opposite side of membrane during dialysis. Solutes in the blood pass through the membrane and into the dialysate due to a concentration gradient.
finger pump	A type of positive displacement pump that sequential compression of the tubing forces compress fluids to move in one direction.
fluid power	The power of the fluid leaving a pump.
heparin	A common blood anticoagulant often used during hemodialysis.
Matlab	Computing software used to solve mathematical problems.
MEMS	Micro ElectroMechanical Systems are very small devices, with parts on the scale of micrometers.
persitaltic pump	A type of positive displacement pump that is used during hemodialysis. Medical tubing is squeezed by rollers to create the pressure gradient.
positive displacement pump	Any type of pump that creates a rise in pressure by cyclic filling and emptying of a volume. Rotary pumps, the pumps most commonly used in hemodialysis, are positive displacement pumps.
recirculation	When dialysate is reused after it has passed through the dialyzer.
Simulink	A commercial system modeling software developed by The MathWorks.

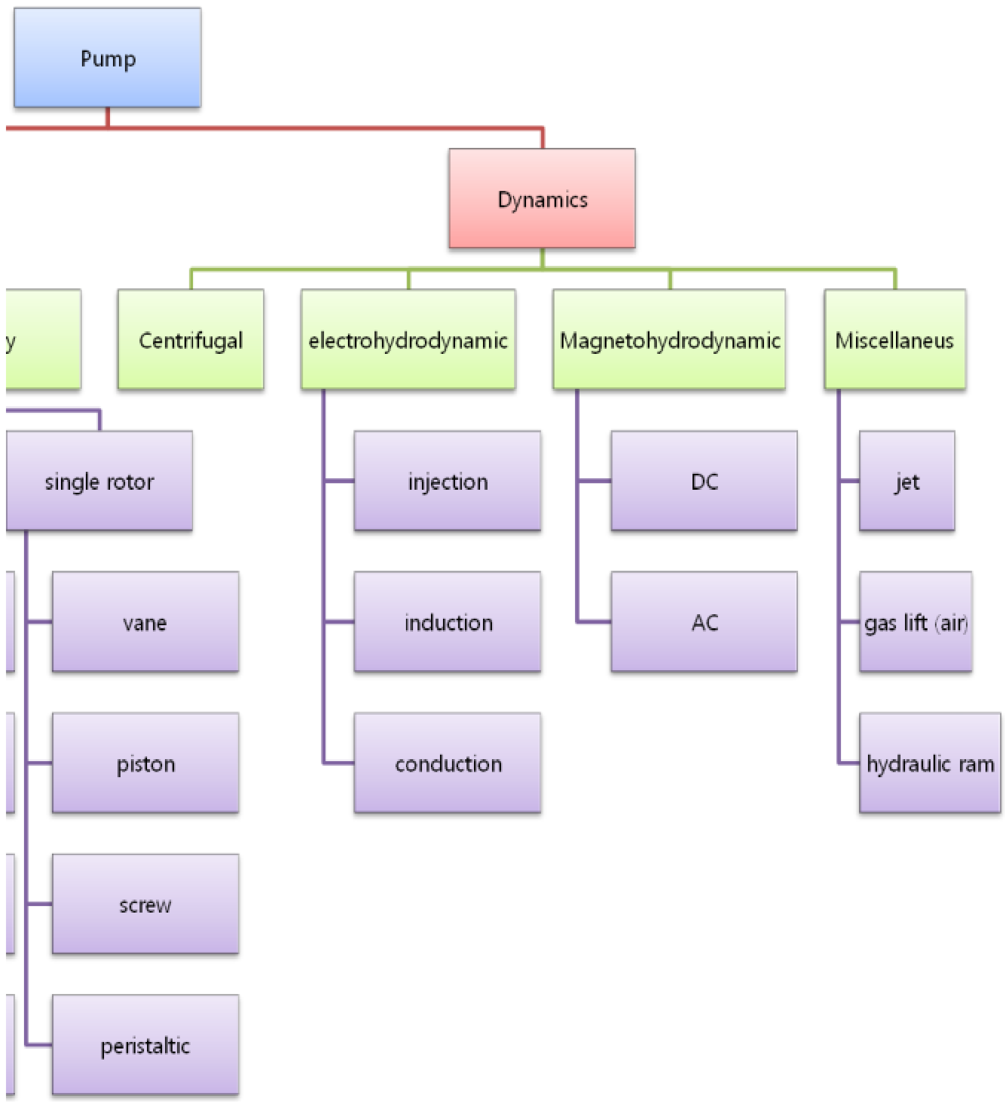
single-pass	When dialysate is passed through the dialyzer just once during hemodialysis.

APPENDIX B

FIGURES



Left half of Figure 3-3. Pump tree PD pumps



Right half of Figure 3-3. Pump tree Dynamic pumps

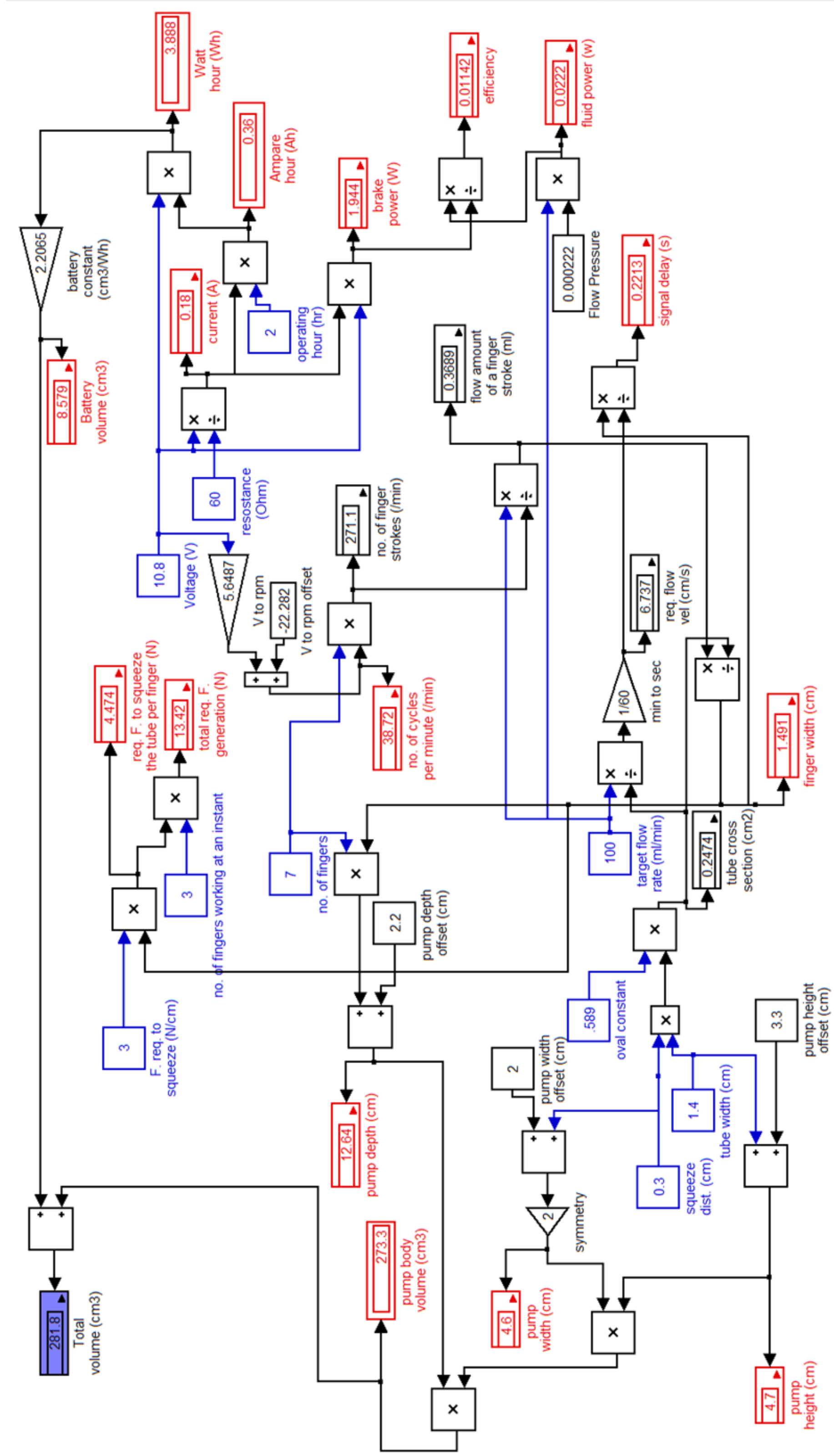
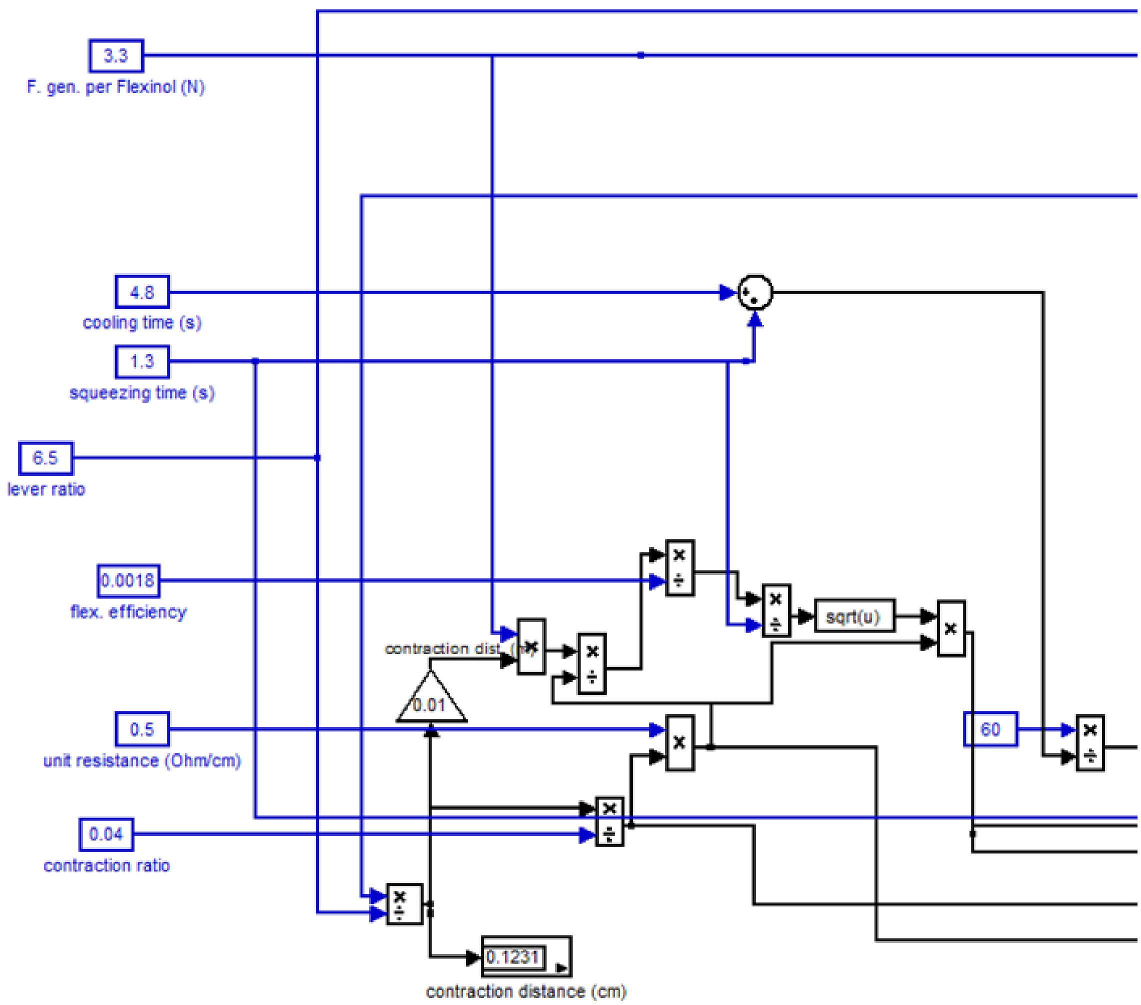
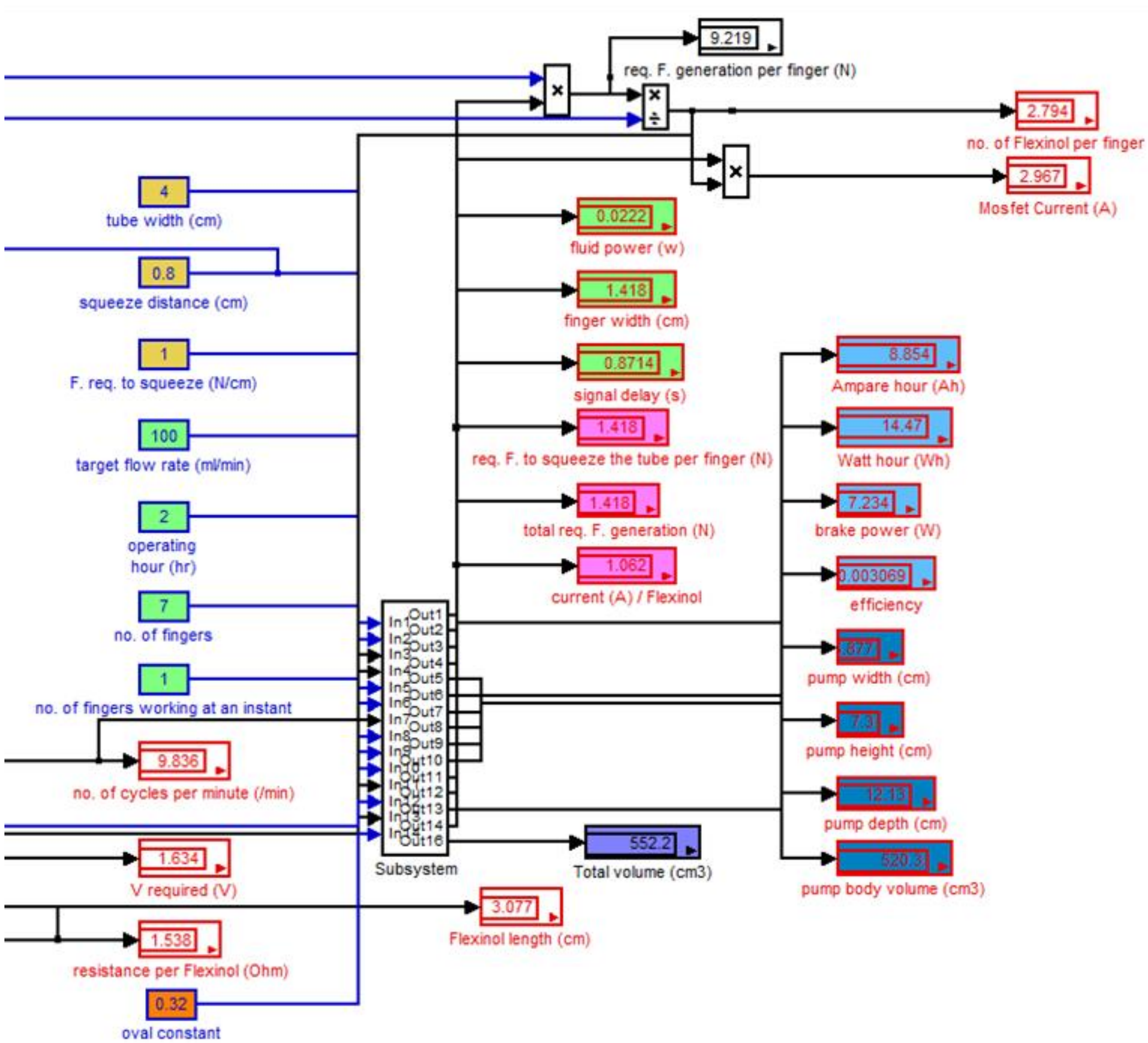


Figure 5-2. Base finger pump model in original form



Left half of Figure 5-4. Adjusted finger pump model in original form



Right half of Figure 5-4. Adjusted finger pump model in original form

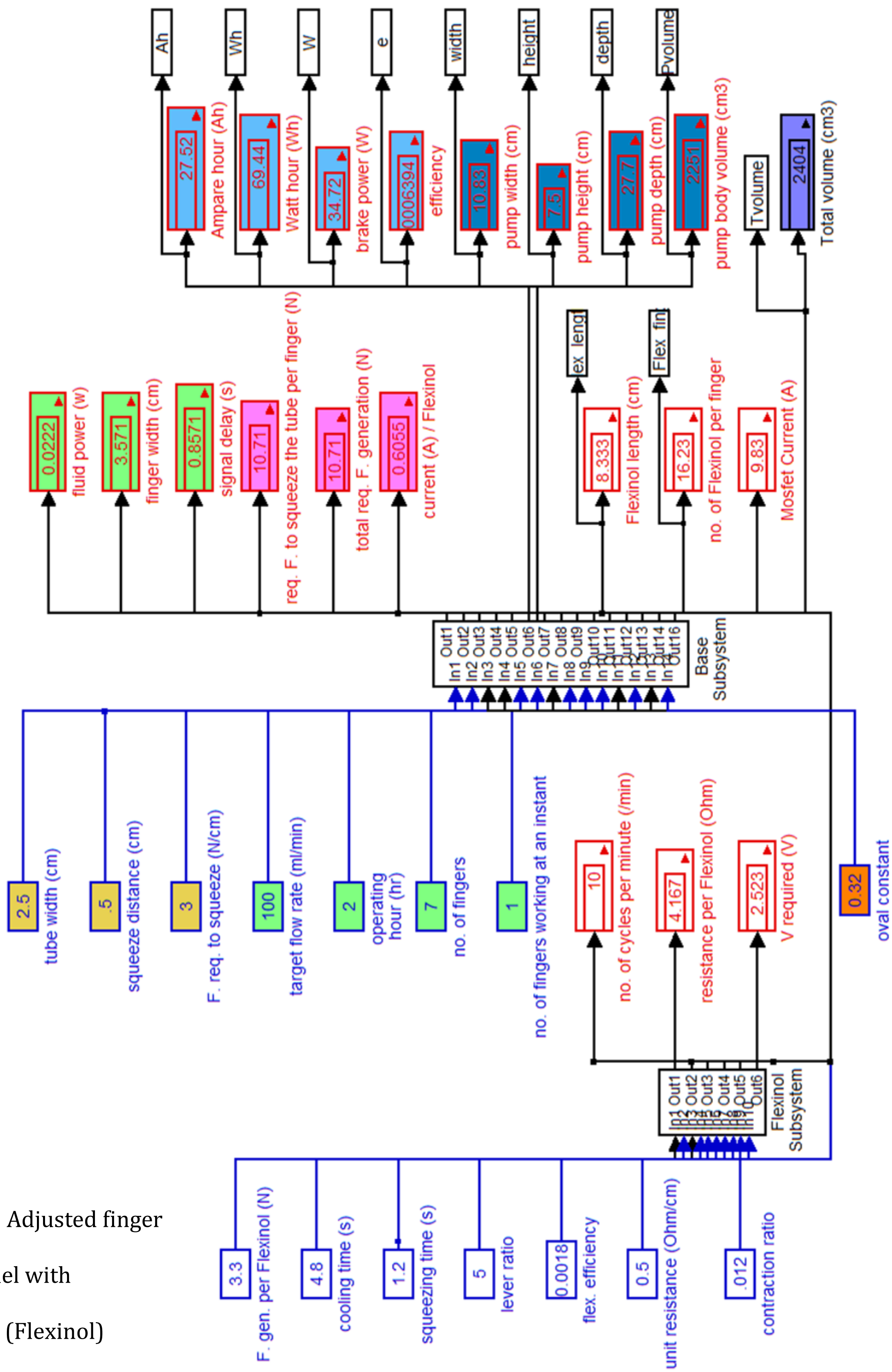


Figure 5-5. Adjusted finger pump model with subsystem (Flexinol)

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