Peristaltic Pump Design for Soft Solid Materials

by

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Thesis Submitted in Partial Fulfillment of the
Requirements for the Degree of
Bachelor of Applied Science

in the
School of Engineering Science
Biomedical Engineering

April 2024
SIMON FRASER UNIVERSITY

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Abstract

This thesis presents the design and initial proof of concept of a peristaltic pump tailored for soft solid materials, focusing on its potential application as part of a prosthetic colon. The research delves into the process of prototype construction, including the selection of materials and design considerations and limitations. A comprehensive examination of the operational cycle highlights the device's functionality, with a particular focus on material propulsion. Challenges encountered throughout the design phase, such as material selection and peristaltic design optimization, are thoroughly discussed. Notably, the study's major accomplishment lies in demonstrating the feasibility of peristaltic pump utilization for soft solid properties.

In a series of eight trials, key performance metrics were established: an average flow rate of 0.39 grams per second, with an average stroke time of 11.82 seconds. The device exhibited an average performance of 7.9%, ejecting an average of 4.7 grams of material per stroke. These findings underscore the device's functionality and lay the groundwork for further refinement and optimization.

Furthermore, the thesis explores potential avenues for future enhancements, including miniaturization, automation, and the quest for implantable medical-grade materials. This work underscores the ongoing efforts to refine the device into a self-sufficient and user-friendly solution, closely aligned with the standards of implantable medical-grade devices.

Dedication

I dedicate this thesis to the resilient individuals who have undergone permanent ostomy surgeries, facing the challenges of gastrointestinal conditions with unwavering strength. May the findings of this research pave the way for future technologies that enhance the quality of life for these patients.

Acknowledgements

I would like to express my deepest gratitude to my supervisor, Dr. Bonnie Gray, for her unwavering support, invaluable guidance, and encouragement throughout the duration of this thesis. Dr. Gray's expertise, patience, and mentorship have been instrumental in shaping my research journey and fostering my academic growth.

I am also immensely thankful to my committee members, Dr. Michael Hegedus and Dr. Ash Parameswaran, for their insightful feedback, constructive criticism, and valuable contributions to this work. Their expertise and guidance have been invaluable in refining my ideas and ensuring the quality of this thesis.

Special appreciation goes to Juan Ferrer for his exceptional guidance and assistance with 3D printing. His expertise and dedication have been indispensable in bringing the technical aspects of this project to fruition.

I am deeply indebted to my family and friends for their unwavering support, understanding, and encouragement throughout this academic endeavor. Their love, encouragement, and belief in me have been my pillars of strength, enabling me to overcome challenges and pursue my goals with determination.

Lastly, I extend my heartfelt thanks to all those who have supported me along this journey. Your encouragement, advice, and presence have made this endeavor not only possible but also immensely rewarding.

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List of Acronyms

MiL	Microinstrumentation Laboratory
CAD	Computer-Aided Design
IV	Intravenous
PDMS	Polydimethylsiloxane
DC	Direct Current
DPDT	Double Pole Double Throw
GI	Gastrointestinal
MPa	Megapascal
SLA	Stereolithography
TPU	Thermoplastic Polyurethane

Glossary

Peristaltic Pump	A pump that operates by compressing a flexible tube to propel fluids or soft solid materials through it, mimicking the peristaltic movement observed in natural processes like digestion.
Colostomy	A surgical procedure involving the diversion of one end of the colon through an opening in the abdominal wall, usually resulting in the use of a bag to collect fecal waste.
Linear Actuator	A device that converts rotational motion into linear motion, often used in mechanical systems to perform tasks such as pushing, pulling, or lifting.
FEP Film	A fluoropolymer film commonly used as a non-stick lining in 3D printers, facilitating the easy removal of printed objects.
Prosthetic Colon	An artificial device designed to replicate the functions of the colon, typically used to restore peristalsis functions of digestive system in individuals who have undergone colonic surgeries.
Biomedical Engineering	A field of study that applies principles of engineering and biological sciences to develop solutions for healthcare and medical challenges.
Miniaturized Device	A small-scale device designed for implantation or use in confined spaces, often used in medical applications for diagnostics, monitoring, or treatment.
Prototype	A preliminary version of a device or system built to test its design, functionality, and performance before mass production or implementation.
Heart-Lung Machine	A device used in cardiac surgery to temporarily take over the function of the heart and lungs during surgery, allowing surgeons to operate on a still heart.
DE - Dielectric Elastomer	A class of electroactive polymers that change shape when exposed to an electric field, often used in actuator applications.
Pulsatile Flow	Flow characterized by periodic fluctuations in pressure or flow rate, as opposed to steady, continuous flow.
Occlusion	The degree to which the tubing of a peristaltic pump is compressed, affecting flow rate and performance.
Viscoelastic	Having both viscous (fluid-like) and elastic (solid-like) properties, characteristic of materials like soft solids and polymers.

Colostomy Bags	External bags worn by patients who have undergone colostomy surgery to collect waste expelled from the digestive system.
Stereolithography	A type of 3D printing technology that uses light-reactive thermoset materials to create solid objects layer by layer.
Shore Hardness	A measure of the resistance of a material to indentation or penetration by a durometer instrument, often used to assess the flexibility or hardness of elastomers.
Silicone Elastomers	Synthetic polymers with rubber-like properties that are commonly used in medical devices due to their biocompatibility and flexibility.
Ratchet Mechanism	A mechanical device that allows motion in only one direction, typically used to prevent backward movement or to provide incremental movement.
Sterilization	The process of killing or removing all forms of microbial life, including bacteria, viruses, and fungi, from an object or surface to prevent infection or contamination.
IdeaMaker	A slicing software used to prepare digital 3D models for 3D printing by converting them into printable layers and optimizing print settings.
Chitubox	A slicing software commonly used for resin-based 3D printing, allowing users to prepare digital models for printing and optimize print settings.
Fecal Matter	Waste material expelled from the digestive tract, consisting of undigested food, bacteria, and other substances.
Tensile Strength	The maximum stress that a material can withstand while being stretched or pulled before breaking or failing.
Biocompatibility	The ability of a material to perform its desired function within the body without causing harmful effects or reactions.
Catheters	Flexible tubes inserted into the body to deliver fluids, medications, or gases to specific locations or to drain fluids or urine from the body.
Abrasion Resistance	The ability of a material to resist wear, rubbing, or scraping when subjected to friction or contact with another surface.
Chemical Inertness	The lack of reactivity or chemical reaction when exposed to other substances or environments.

Peristalsis	The involuntary constriction and relaxation of muscles in the gastrointestinal tract, creating wave-like movements that propel contents forward.
Implantable Medical Device	A medical device designed to be implanted into the body, often for therapeutic or diagnostic purposes.
Medical-Grade Materials	Materials that meet specific standards for use in medical devices, ensuring safety, biocompatibility, and reliability.

Chapter 1. Introduction

1.1. Overview

Peristaltic pumps, also known as roller pumps, are widely employed in fluidic and microfluidic applications within laboratory settings. These pumps, typically recognized for their utility in pumping a variety of fluids, operate through a unique mechanism involving the compression of a flexible tube enclosed within a circular pump casing [1]. While the majority of peristaltic pumps are designed to function based on rotary motion, it is worth noting that alternative configurations, such as linear peristaltic pumps, have also been developed, for example, the pump design shown in **Figure 1**. However, despite related peristalsis forming the basis for movement of mixed matter in the human colon, it is a notable fact that peristaltic pumps have not been explored in the context of pumping soft solid materials, such as materials that have clay like texture. This limitation opens up intriguing possibilities for applications in the medical field, with one of the most promising prospects being the creation of a prosthetic colon.

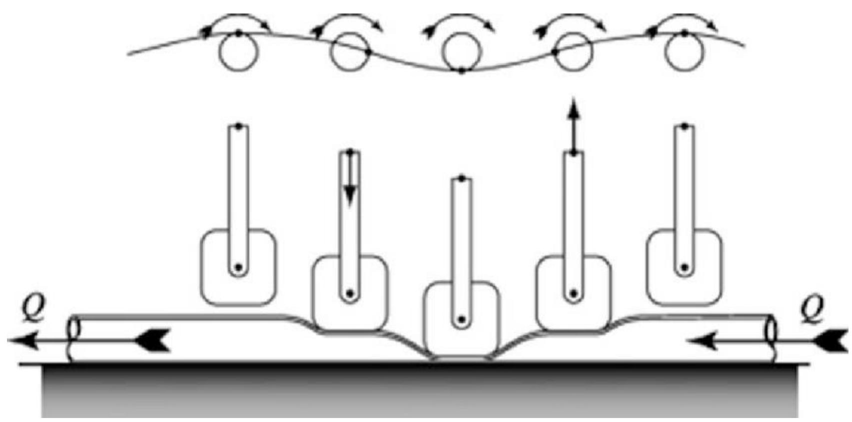

Figure 1: Linear Peristaltic Pump [2]

In medical scenarios where patients require procedures like partial colectomy, which entail the surgical removal of a portion of their colon due to diseases or abdominal injuries, there arises a unique challenge related to the restoration of digestive functions [3]. In such situations, patients undergo colostomy surgeries, a procedure in which a permanent bag is affixed to the outside of their abdominal wall for the collection of fecal waste as shown in **Figure 2**. This procedure significantly impacts the patients' quality of life, as the bag often leads to discomfort, skin irritation around the stoma, and, in certain cases, can even result in blockages. The concept of utilizing a biodegradable material to construct a tube that can serve as a prosthetic colon is a potential promising avenue of research. The potential benefits of such an innovation are manifold, especially in the context of maintaining and restoring the essential physiological process of peristalsis, which is responsible for bowel movements.

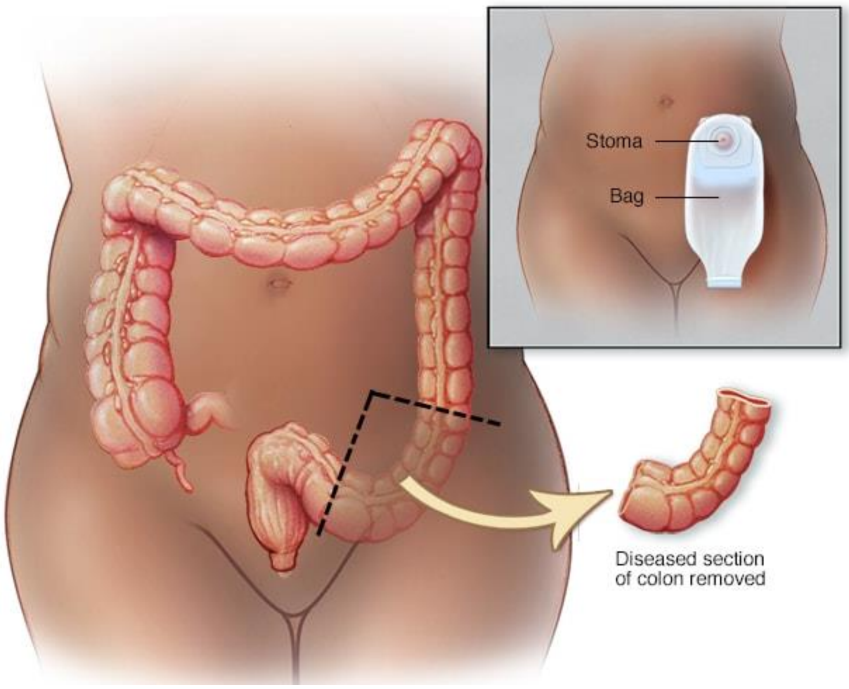

Figure 2: Colostomy Procedure and Ostomy Bag [4]

This thesis centers on the exploration and experimentation of the peristaltic movement applied to soft solid materials, specifically targeting its relevance in medical contexts. The proposed system comprises essential components, including a linear actuator, a series of movable 3D-printed blocks, a structural framework for device integration, and a flexible tube. The research involves the introduction of a clay-like material into the tube to assess and test the system's functionality.

In operation, the 3D-printed blocks are strategically placed on top of the flexible tube. As the linear actuator extends from one side, it sequentially applies force to each block, compressing the flexible tube and facilitating the forward movement of the soft solid material. Upon reaching full extension, the linear actuator retracts to the initial position. Simultaneously, the blocks retract upwards with the assistance of elasticity of the tube, allowing the flexible tube to expand to its original diameter. This cycle readies the system for subsequent movements, accommodating the arrival of new soft material.

The fundamental aim of this mechanism is to replicate the peristaltic movement critical for effective bowel functioning, particularly in sections of the colon that have undergone surgical removal. Essentially, the system emulates the wave-like motion inherent in peristalsis, crucial for propelling materials through the digestive tract.

This thesis aims to delve into the theoretical and practical aspects of this application of peristaltic pump technology, specifically for the movement of soft solid material for medical purposes, through development of a simple proof-of-concept mechanism. By successfully developing and implementing this system, it opens new doors to advancements in the field of medical devices, ultimately contributing to the well-being and quality of life for individuals who have undergone colon-related surgeries such as partial colonic resection.

1.2. Contributions

This project originated as an undergraduate research initiative within the Microinstrumentation Laboratory (MiL), and it has enhanced the lab's expertise in the domain of peristaltic pump design for soft solid materials. The contributions made by this thesis to MiL's capabilities and the field of peristaltic pumping are outlined as follows:

1. Exploring the development of 3D-printed soft and flexible tubes that not only exhibit flexibility but also possess structural integrity. This dual characteristic is crucial in mimicking the complex nature of the human colon.

2. Design of a proof-of-concept peristaltic mechanism that replicates human bowel movements. This mechanism demonstrates the transfer of fecal matter within a tube, emulating the intricate peristaltic motion observed in the natural digestive process.

3. Proposing prosthetic colon to extend its impact by presenting novel methods and designs that pave the way for the eventual construction of a prosthetic colon. These suggestions contribute valuable insights and directions for the MiL and other research groups to explore in the realm of biomedical

engineering, offering potential advancements in medical prosthetics and therapeutic devices.

1.3. Thesis Goal

The primary objective of this thesis is to delve into the intricate dynamics of the peristaltic movement, exploring its potential applications within the medical industry, specifically as a fundamental component of a prosthetic colon. It is imperative to emphasize that the focus of this research is not directed towards developing a fully miniaturized device suitable for human implantation. On the contrary, the prototype designed for this investigation intentionally utilizes non-biodegradable materials and is intentionally upscaled. This deliberate choice enables a comprehensive study of the peristaltic movement's effectiveness on soft solid materials within a flexible tube.

In essence, the goal of this thesis is to research, design, and develop a proof-of-concept prototype device capable of executing the peristaltic movement. The emphasis lies in creating a functional apparatus adept at transferring soft solid materials from one end of a tube to the other. This approach allows for a detailed examination of the peristaltic mechanism's efficacy in facilitating the movement of substances within a controlled environment, laying the foundation for potential advancements in medical applications.

1.4. Thesis Organization

This thesis is organized into five comprehensive chapters, each contributing to an understanding of the research undertaken. Chapter 1 serves as an introduction, presenting the identified problem and elucidating the proposed system's key components. This chapter lays the foundation for the subsequent chapters by outlining the context and purpose of the research.

Chapter 2 delves into the background of the study, offering a thorough exploration of current research in the field and an examination of various peristaltic pumps. This chapter provides essential contextual information to situate the proposed system within the broader landscape of peristaltic movement studies.

Chapter 3 constitutes a deep dive into the design of the system. It details the development of flexible tubes, the proposed peristaltic pump mechanism, and conducts a comparative analysis with other designs. The chapter also expounds on the methodology employed in the research, elucidates the materials chosen for prototype development, and showcases the Computer-Aided Design (CAD) representations of the device's components. Additionally, it furnishes specifications of the linear actuator incorporated into the device.

Moving forward, Chapter 4 discusses the design and assembly process, presenting the finalized prototype device. This chapter suggests suitable fecal substitutes for testing, outlines the testing procedures and setup, and systematically reports the results obtained. A subsequent discussion provides valuable insights into the outcomes, enriching the reader's comprehension.

Finally, Chapter 5 encapsulates the thesis by addressing challenges encountered, proposing avenues for future work, and offering conclusions. This chapter serves as a crucial reflection on the research journey, emphasizing the contributions made and paving the way for potential further advancements in the field.

Chapter 2. Background and Methodology

The inception of peristaltic pumps traces back to the mid-19th century when Porter and Bradley secured the patent for the first roller pump in 1855, initially operated by hand. Subsequent advancements led to the creation of a modified version called the "surgical pump" by E. E. Allen in 1887, tailored for direct blood transfusion. The evolution continued with Truax introducing the first double roller pump in 1899, building on the Allen pump's foundation. Over the ensuing decades, luminaries such as Beck, Van Allen, Bayliss, Müller, Henry, and Jouvelet refined the apparatus, advocating roller pumps for various applications, notably blood transfusion.

DeBakey's 1934 modifications, coupled with their utilization in one of the initial heart–lung machines by Gibbon, cemented the association between DeBakey and this type of pump. In 1935, for perfusion experiments, Fleisch pioneered the use of an electrically powered roller pump. Presently, the roller pump stands as the predominant choice for cardiopulmonary bypass globally, triumphing over early pulsatile tube compression pumps and ventricular pumps [5].

Peristaltic pumps play a pivotal role in pumping fluids that require a sterile or contamination-free environment, making them particularly valuable in applications involving clean or highly reactive substances. This characteristic holds significant relevance for prosthetic colons, where the utmost priority is to prevent contact with fecal matter to mitigate the risk of contamination and infection. Given that a prosthetic colon typically replaces a surgically removed section of the colon and is intricately connected at both ends to the existing colon, the pumping mechanism is crucially important. To maintain the required hygiene standards, it is imperative that the peristaltic pump operates externally to the prosthetic tube, mirroring the functionality of conventional peristaltic pumps. This design ensures that the pumping action is performed from the outside of the tube, aligning with the specific needs and intricacies of prosthetic colons, and emphasizing the importance of maintaining a sterile environment in such medical devices [6], [7].

Common applications of peristaltic pumps encompass the administration of IV fluids through infusion devices, apheresis, handling highly reactive chemicals, managing high-solids slurries, and other scenarios necessitating stringent isolation from the environment. Their deployment in heart–lung machines during bypass surgery and hemodialysis systems is attributed to their minimal impact on hemolysis and blood cell rupture. Additionally, other applications of peristaltic pumps include utilization in hemodialysis machines, medical infusion pumps, pharmaceutical production, and chemical analytical equipment [6].

Typically, peristaltic pumps exist in two main types: circular or rolling pumps and linear peristaltic pumps. Chapter 2 of this thesis will delve into prior research specifically focusing on peristaltic pumps for soft solid materials and expound on the mechanisms underlying these two types of peristaltic pumps.

2.1. Previous Research

The research by Carpi et. al. focuses on the development of a lightweight and deformable structure as presented in **Figure 3**, with intrinsic distributed electromechanical actuation, specifically designed for soft linear peristaltic pumps intended for incompressible fluids. The system comprises radially expanding flexible tubular modules constructed from dielectric elastomers (DEs), a class of electroactive polymers. Each module, consisting of a cylindrical hollow DE actuator, operates in a purely radial mode with defined boundary constraints.

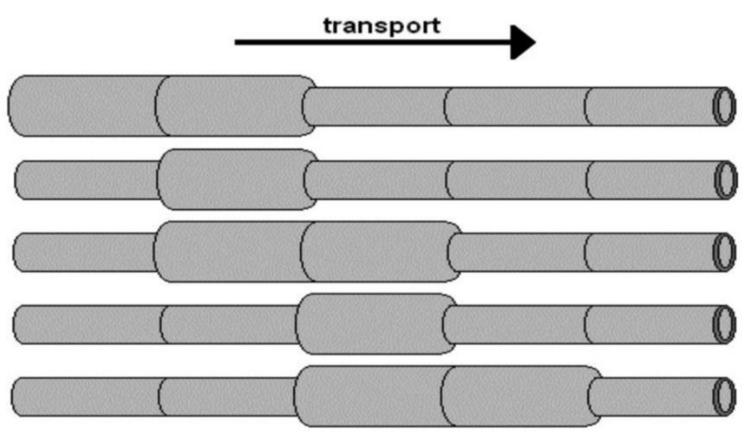

Figure 3: Radially Expanding Modules In a Linear Peristaltic Pump [7]

This innovative system is motivated by the need for fluid transport systems addressing specific requirements such as low complexity, high mechanical compliance, power-to-weight and power-to-volume ratios, response speed, long lifetime, electrical driving capabilities, and ease of scaling. The proposed concept involves a flexible tubular structure with embedded distributed actuation, enabling electrically controllable dimensional changes. This configuration makes the system suitable for a range of applications, mainly for fluid transportation.

The study explores the electromechanical transduction performance of these DE modules through analytical, numerical, and experimental investigations. The results provide insights into predicting system performance concerning displaced volume and driving pressure, considering factors such as material elastic modulus and applied voltage [7].

While the research presents an ideal mechanism for fluid transport, it is essential to note that it is based on liquid pumping, and there is no explicit mention of soft solid materials. Despite this limitation, the innovative approach to DE actuation and the potential for a flexible structure with distributed actuation make it an intriguing concept for exploration in the context of colon prosthetic applications. Further research and adaptation may be necessary to address the specific challenges associated with soft solid material transport, as distinct from incompressible fluids.

Currently, there is a noticeable gap in research pertaining to the application of peristaltic pumps for colon prosthetics. Despite their established biomedical applications, notably in the precise injection of intravenous (IV) fluids, the exploration of peristaltic pumps for colon prosthetics remains largely uncharted territory.

Furthermore, a similar research void is observed in the context of using peristaltic pumps for handling viscoelastic substances and soft solid materials. Most academic literature predominantly concentrates on the transfer of fluids and liquid materials through peristaltic pumps, leaving a knowledge gap in understanding the dynamics of peristaltic pumping concerning materials with viscoelastic properties.

While the use of peristaltic pumps for fluid transfer is well-documented, there is an evident need for expanded research endeavors encompassing diverse material properties, including viscoelastic substances and soft solids. Such investigations could unlock innovative applications of peristaltic pumping technology, potentially revolutionizing fields such as gastrointestinal prosthetics and material handling in biomedical engineering. Researchers are encouraged to delve into these unexplored realms, addressing the current limitations and paving the way for advancements in the utilization of peristaltic pumps across a broader spectrum of materials and applications.

2.2. Current Peristaltic Pumps

The optimal peristaltic pump would feature an infinitely large pump head diameter and rollers of the utmost diameter, ensuring an extended tubing lifespan and a consistent, pulsation-free flow rate. While achieving such an ideal peristaltic pump remains impractical in reality, careful design considerations can bring these parameters closer to the ideal, enhancing performance. Such careful design considerations can provide extended tubing longevity and sustaining a stable, accurate flow rate for several weeks without the risk of tubing rupture.

Peristaltic pumps come in various types, with rotary and linear peristaltic pumps standing out as the primary categories. In the ensuing subsections, a detailed exploration of these two peristaltic pump types will shed light on their distinctive features and applications, contributing to a comprehensive understanding of their functionalities in various settings.

2.2.1. Rotary Peristaltic Pumps

The rotary peristaltic pump operates through the successive compression of a flexible hose by two or more pressing shoes assembled on a rotating wheel, as depicted in **Figure 4**. The first shoe initiates the process by creating a vacuum through the compression of the hose walls, drawing the pumped liquid into the hose. Once the liquid enters the hose, the second roller comes into play, pushing the liquid towards the pump outlet.

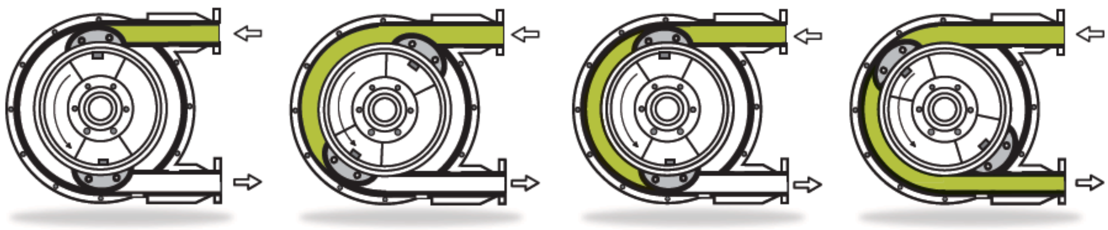

Figure 4: Rotary Peristaltic Pump Flow Scheme [9]

As the shoes continue to rotate, the previously compressed section of the hose begins to rebound to its nominal state, generating a vacuum that pulls more fluid into the hose. The strategic detachment of the shoe at the discharge side, while the opposing shoe is already compressed, prevents any backflow. This ensures the efficient pulling in and pushing out of the pumped fluid as a result of the wheel rotation. Components of a rotary peristaltic pump include the pump case, hose, rotor, shoe, and motor. The pump case houses the rotor, supported by bearings on a shaft. Shoes are typically placed at the rotor, and flanges connect input and output fluids [7], [8], [9].

Differentiating itself from the linear type, which will be discussed in the subsequent section, the rotary peristaltic pump utilizes a motor for shaft rotation, while the linear type relies on a cam mechanism. The rotary design proves advantageous for prosthetic colons, especially in sections with non-linear configurations, such as the Hepatic Flexure and Splenic Flexure, as illustrated in **Figure 5**.

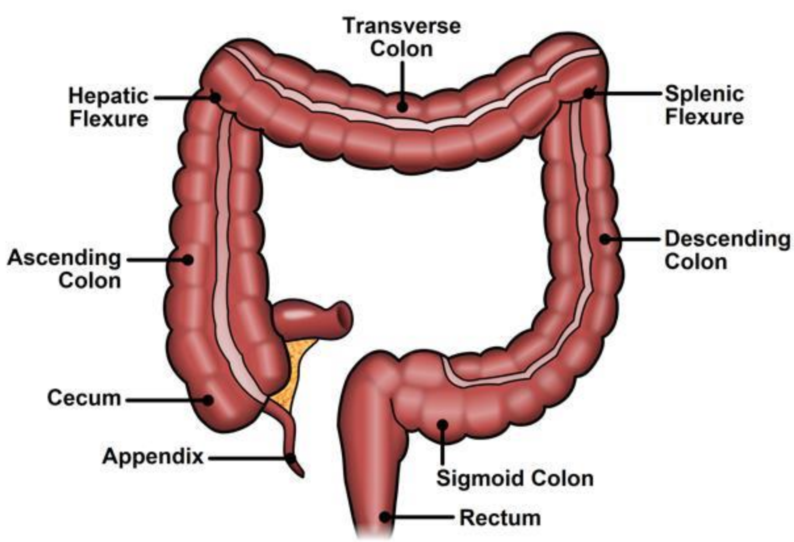

Figure 5: Large Intestine Anatomy [10]

2.2.2. Linear Peristaltic Pumps

In a study conducted by Xiang et al. [11], an elucidation of a linear peristaltic pump design tailored for microfluidics is presented. The mechanism comprises essential components, including 3 cams, a camshaft, 3 followers, a PDMS tube microchannel, and a substrate, as illustrated in **Figure 6**. The intricacy of this design becomes apparent as the camshaft, set into motion by a motor, rotates the three cams in a synchronized manner. This specific synchronization ensures that the cam's nose applies pressure to the microfluidic channel, sequentially squeezing the tube to propel the sample forward. The cyclic motion of this process is depicted in **Figure 7**.

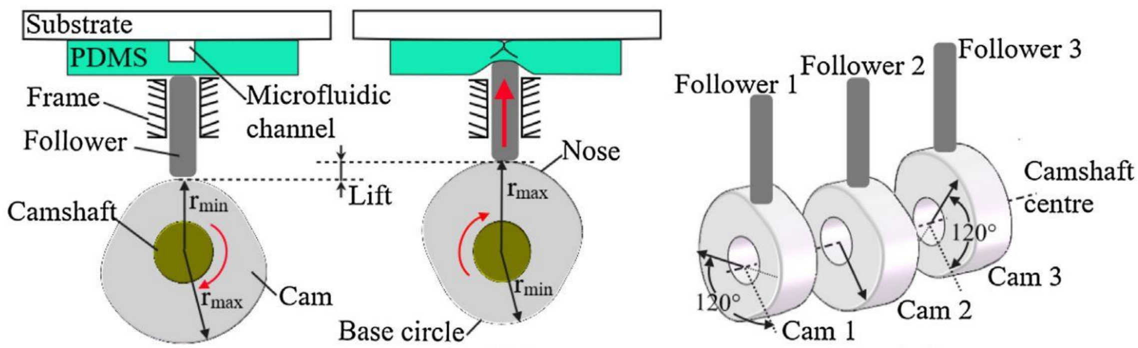

Figure 6: Schematic Diagram of A Linear Peristaltic Pump Mechanism [11]

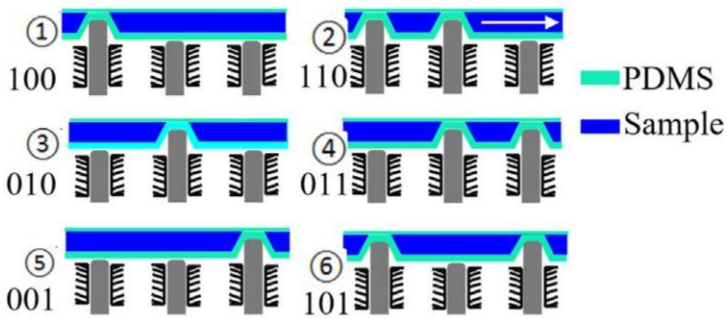

Figure 7: A diagram of a linear peristaltic pump showing the movement of the sample through a PDMS channel [11]

This innovative approach draws on historical applications of cam-follower structures in engine valve timing, underscoring the simplicity and stability inherent in controlling actuation time sequences. The study introduces rotating cams characterized by a dwell-rise-dwell-return profile, which actuates followers in a sequential manner, systematically compressing the microfluidic channel for effective peristaltic pumping.

Furthermore, the paper highlights the adaptability of this mechanism, allowing for a wide range of follower displacements, and emphasizes the control of the refreshing frequency through the manipulation of cam rotation. The integration of rotating cams in this study represents a significant advancement, introducing a nuanced approach to the sequential compression of the microfluidic channel, a crucial element in the functionality of peristaltic pumps.

2.3. Key Specifications and Factors in Peristaltic Pump Design

Peristaltic pumps are subject to specifications, with occlusion being one of the parameters. Occlusion, determining the amount of squeeze applied to the tubing, is expressed as a percentage of twice the wall thickness or as an absolute amount of the wall that is squeezed. Typically ranging from 10% to 20%, occlusion varies based on tube material hardness [12]. It is worth noting that occlusion is not an important factor in designing prosthetic colon, since the material being pumped is soft solid and will significantly affect the value of occlusion.

Additionally flow rate in peristaltic pump plays an important factor. The flow rate is a pivotal parameter influenced by factors such as tube inner diameter, pump-head outer diameter, pump-head rotational speed, and inlet pulsation. The number of rollers, surprisingly, does not increase the flow rate but affects the pulsing amplitude at the outlet. Longer tubes, although not impacting the flow rate directly, increase pressure generation by introducing more pinch points [13].

Advantages of peristaltic pumps include contamination prevention due to the isolated contact of the content with the tube interior only. Additionally, their low maintenance requirements, and ability to handle various fluids make them cost-effective. Peristaltic pumps prevent backflow without valves and provide a consistent fluid amount per rotation [6]. Backflow in small amounts is normal in human colon [14], thus in this case minor backflow is not an important factor for the proposed design. However, drawbacks of peristaltic pumps include the degradation of flexible tubing over time which can cause leakage, and limitations in effectiveness with high liquid viscosity [6].

In comparing rotary and linear peristaltic pumps, each design possesses its unique applications and advantages. Rotary peristaltic pumps, due to their typically bulkier nature, are less conducive to miniaturization [15], making them less suitable for prosthetic applications. On the other hand, linear peristaltic pumps offer a simpler construction process compared to their rotary counterparts. Given this evaluation, the decision has been made to utilize a linear peristaltic pump design for the prototype. The subsequent chapter, Chapter 3, will provide an in-depth exploration of the prototype's design.

Chapter 3. Prototype Design

In this chapter, we delve into the details of the prototype device designed specifically to validate the feasibility of peristaltic movement for soft solid materials. It is crucial to emphasize that the primary objective of this prototype is to confirm the viability of peristaltic motion for such materials, rather than serving as the basis for a prosthetic colon. Consequently, the materials utilized in the prototype are non-biodegradable, and the device's size has been deliberately scaled up, deviating from proportions suitable for a prosthetic colon.

The chapter describes a system overview, providing comprehensive insights into the design, the materials chosen for constructing the tube, and the overall device structure. Subsequently, the complexities of the peristaltic movement design are examined. Moreover, this chapter encompasses a thorough discussion of alternative materials and designs, facilitating a comparative analysis. Additionally, it will address the limitations and constraints inherent in the prototype, offering a holistic understanding of its scope and potential applications.

3.1. System Overview

The proposed prototype device comprises designed and selected components to facilitate the peristaltic movement of soft solid materials, as discussed in subsequent sections outlining design specifics and material choices. The system overview introduces the various parts of the device. First, a flexible tube acts as a duct to facilitate material passage. To pump materials from one end of the tube to the other, a mechanism is employed, consisting of a base, triangular blocks, and a linear actuator. In its relaxed state, the triangular blocks rest atop the tube. When the linear actuator applies force to the hypotenuse of the triangular blocks, they move downward, compressing the tube and propelling materials forward. Upon retraction of the linear actuator, the tube return to a relaxed state, ready for the next set of materials and peristaltic movement.

Clay with the desired consistency has been selected for testing the device's durability and reliability. **Figure 8** illustrates the final assembled prototype device.

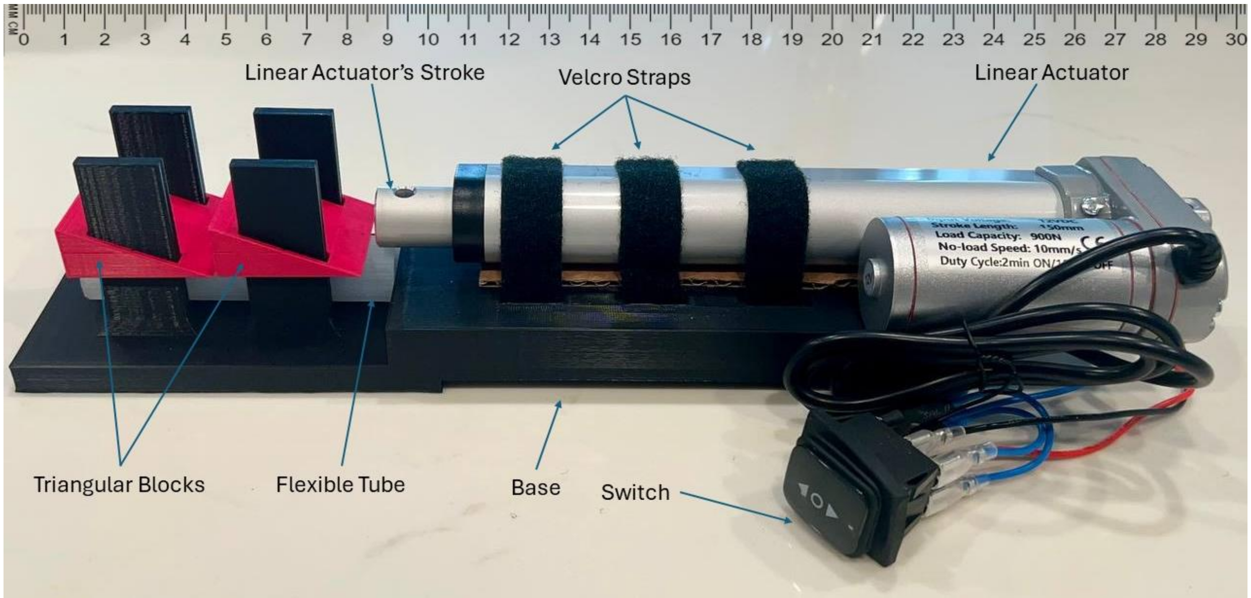

Figure 8: Final Assembled Device with Labels and a Sclae Bar in CM

3.2. Flexible Tube Development

In this section, a comprehensive exploration of the development of the flexible tube for the prototype is undertaken. Before delving into the technical aspects, it is essential to elucidate the imperative need for such a tube.

Colectomy, a surgical procedure involving the partial or complete removal of the patient's colon, is often necessitated by conditions like cancerous tumors or Crohn's disease. Following the repair or removal of the colon, surgeons undertake the crucial task of reconnecting the digestive system to facilitate waste expulsion. This reconnection can take various forms, including the rejoining of remaining colon portions or the creation of an opening in the abdomen, known as colostomy. Patients who undergo colostomy are required to wear a bag externally, serving as a receptacle for waste [2], [16]. Unfortunately, colostomy bags pose significant discomfort for patients, negatively impacting their quality of life and leading to issues such as skin irritation, infection, and potential stoma blockages [16], [17].

In the envisioned prototype, the flexible tube assumes the pivotal role of the colon, serving as a conduit for waste passage from one end of the dissected colon to the other. This design aims to offer an alternative to conventional colostomy bags, addressing the discomfort and complications associated with their use. The subsequent exploration will unravel the intricacies of tube design, including the choice of materials. Additionally, alternative tube designs are presented and compared, providing a comprehensive understanding of the possibilities in enhancing patient comfort and well-being.

3.2.1. Flexible Tube Design

To devise the optimal design for the flexible tube, a thorough understanding of the anatomy of the large intestine is imperative, given that the tube aims to replicate certain aspects of this anatomical structure. **Figure 4** provides an overview of the general anatomy of the human large intestine, highlighting its seven distinct regions. The cecum serves as a dead-end pouch, featuring the appendix as a small finger-like projection at its ventral end. Fecal matter travels from the cecum upward through the ascending colon, horizontally across via the transverse colon, and then descends through the descending colon and sigmoid colon. The rectum, the terminal section of the large intestine, is then connected to the anus for defecation [18].

Research conducted by Helander et al. underscores the variability in the diameter of the human large intestine, averaging around 4.8 cm [19]. Different regions within the large intestine exhibit distinct diameters, and inter-individual variations further contribute to this diversity. Furthermore, the diameter of the colon undergoes dynamic changes during peristaltic movement. In the context of a prosthetic colon, customization of the tube dimensions becomes imperative to align with the unique anatomy of each individual.

In the envisioned system, a deliberate decision was made to design a flexible tube with an outer diameter of 2 cm, an inner diameter of 1.8 cm, and a thickness of 1 mm. The tube's length is set at 10 cm, providing ample space for the study of material movement within the tube. **Figure 9** illustrates the CAD representation of the tube. The subsequent section delves into a comprehensive discussion of the chosen material for the tube, shedding light on its selection criteria and implications for the prototype's functionality.

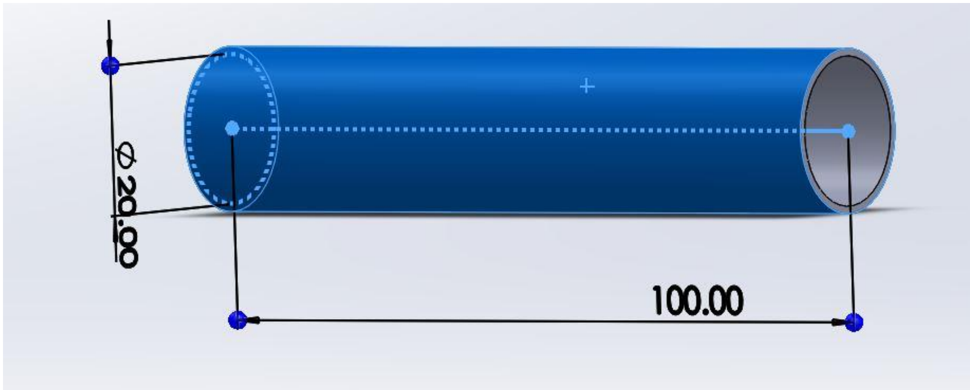

Figure 9: CAD Desing of the Tube – Measuremnts are in mm

3.2.2. Flexible Tube Material

During the iterative process of developing the flexible tube, various materials underwent testing to determine their suitability for the intended purpose. Initially, a Tygon tube with a 2cm diameter was selected. However, issues arose with Tygon, primarily related to its thickness, measuring around 2mm. This excessive thickness hindered flexibility, making it challenging for the peristaltic mechanism to exert sufficient pressure on the tube to fully squeeze the tube. Additionally, Tygon exhibited high resistance to soft solid materials, impeding their smooth passage through the tube.

Another material subjected to testing was Raise 3D TPU (Thermoplastic Polyurethane), an elastic and flexible filament for 3D printing through the extrusion method. The specifications of TPU are detailed in Table 1. Despite its high tensile strength of 29.3 ± 2.8 MPa and an impressive elongation at break of approximately 330%, the TPU tube faced limitations [20]. The low resolution resulting from the extrusion method compromised the tube's cylindrical shape, leading to holes that rendered it unsuitable for this application—its inability to retain soft solid materials without leakage deemed it impractical.

Exploring an alternative approach, flexible resin through stereolithography (SLA) 3D printing was considered. Shore hardness emerged as a pivotal property, with flexibility being a key criterion for achieving a seamless peristaltic movement similar to the human colon. The Shore hardness scale, ranging from 00A to 100A (with 00A being the softest material and 100A being the hardest) became a decisive factor in material selection. Ultimately, resin Resione F80 was chosen for the prototype due to its shore hardness of 50A, providing optimal flexibility for the application. Despite other Resione's resins like F39/F69 offering superior tensile and tear strength, F80's adequate strength made it the preferred choice for the envisioned prototype design [20].

Material	Shore Hardness	Tear strength	Tensile Strength	Elongation at Break
TPU	95A	N/A	29.3 MPa	330%
F39/69	60~75A	47.2 KN/m	7.9 MPa	255.10%
F80	50A	9.75 KN/m	3.8 MPa	159%

Table 1: Flexible Material Properties [21],[22]

3.2.3. Flexible Tube Methodology

Following the design phase, the CAD design of the tube undergoes slicing using appropriate software such as Chitubox, where essential settings and supports are added. Detailed instructions for settings and support can be found in Appendices A-C. Subsequently, the tube is ready for 3D printing, employing a resin printer. In the case of this prototype, the Elegoo Saturn 8K is utilized.

Resin 3D printing operates through stereolithography (SLA), falling within the vat photopolymerization family of additive manufacturing technologies, commonly referred to as resin 3D printing. These printers operate on a shared principle, utilizing a light source, such as a laser or projector, to solidify liquid resin into hardened plastic. SLA 3D printers specifically utilize light-reactive thermoset materials, known as "resin." When exposed to specific wavelengths of light, these SLA resins undergo polymerization, wherein short molecular chains join together, forming solidified, rigid, or flexible geometries [23].

One notable concern associated with the design of this particular flexible tube is the potential degradation of the tubing over time, leading to the risk of tears and subsequent material leakage. The degradation of the tube can be assessed by measuring its initial thickness and comparing it with the thickness after conducting numerous trials to determine the extent of degradation. This data is essential for evaluating the reliability of such materials. Additionally, other potential risks may include variations in material properties over the course of usage, impact on flexibility due to external factors, and compatibility issues with the intended application. Evaluating and addressing these risks is crucial for ensuring the long-term functionality and reliability of the flexible tube design.

3.2.4. Flexible Tube Alternative

Several techniques and materials can be employed in the construction of flexible tubes, ranging from medically graded materials to alternative flexible substances like silicone.

One notable example of a medically graded material used in a medical device is the Gelweave, produced by the Terumo Aortic company, serving as a graft for aortic prosthesis. This prosthetic incorporates a flexible tube that is constructed from thin, pliable, woven polyester sourced from Vascutek, a global medical device company specializing in solutions for aortic and peripheral vascular disease. The tube, depicted in **Figure 10**, is crafted from Dacron, a type of polyester known for its strength, durability, and abrasion resistance. Gelsoft Plus devices, categorized as fabric tubes or grafts, are another example. These devices feature knitted polyester fabric coated with gelatin and are designed for repairing diseased segments of blood vessels within the vascular system. Dacron polyester fabrics, renowned for their robustness, make these tubes available in various sizes, ranging from 6 to 38 mm [24],[25],[26].

Figure 10: Aortic Root Tube by Gelweave [25]

While these medically graded tubes present an ideal alternative for a flexible tube due to their material characteristics and flexibility, their limited commercial availability renders them impractical for use in the proposed prototype. Despite their unavailability for this specific application, exploring such alternatives provides valuable insights into potential advancements in flexible tube technology for prosthetic colon applications.

Silicone, another material extensively employed in implantable medical devices, offers remarkable versatility, allowing for diverse shapes and flexibility. This material presents several advantages, including superior biocompatibility and hypoallergenic characteristics, chemical inertness with resistance to oil, solvents, and stains, bacterial resistance, easy cleaning and sterilization, and a prolonged shelf life while maintaining performance in harsh conditions (-80°C to 300°C). Since the late 1960s, silicone elastomers have been integral to long-term implants, playing a role in joint arthroplasty, catheters, drains, and hydrocephalus shunts. Demonstrating unwavering safety over half a century, silicone elastomers have empowered millions of patients to overcome debilitating conditions [27].

Notably, silicone rubber exhibits high flexibility and resilience, returning to its original shape after flexing or bending, as indicated by a MatWebb silicone rubber data sheet reporting a shore hardness of approximately 95A [28]. In contrast, Rsione F80 resin, with a shore hardness of 50A, surpasses silicone in flexibility. Additionally, the convenience of modifying and customizing designs using a 3D printer makes the resin flexible tube the preferred choice for the prototype design.

3.3. Peristaltic Movement Mechanism

In the preceding sections, we provided an overview of peristaltic pumps, elucidating their operational principles and associated advantages. In the context of a prosthetic colon, the need arises for a mechanism capable of propelling feces through a tube connected to the existing colon from both ends. As feces enter the flexible tube, previously designed and detailed, a mechanism is essential to facilitate their forward movement, ensuring delivery to the opposite end of the tube, where the patient's colon can take over the natural digestive process.

Human peristalsis, driven by contractions of soft muscles within the GI tract, serves as a natural model for this mechanism. Similarly, a prosthetic colon necessitates an external mechanism to emulate peristaltic movement, transferring feces efficiently. To prevent infection and cross-contamination, this mechanism must operate externally to the tube. However, the current peristaltic pumps commercially available are only used for liquid materials and they are only equipped with small diameter tubes, which limits experimentation with soft solid materials. As previously mentioned, the diameter of the tube significantly affects the flow rate. Therefore, there is a critical need to design a peristaltic pump that can utilize larger diameter tubing to enable experimentation and application with soft solid materials. Drawing inspiration from traditional peristaltic movement, the subsequent sections delve into the proposed triangular peristaltic mechanism design and methodology. Additionally, alternative designs will be explored and compared for a comprehensive understanding.

3.3.1. Triangular Block Peristaltic Mechanism Design

The triangular block peristaltic mechanism comprises fundamental elements: a base, two triangular blocks, and the tube, with the linear actuator serving as a crucial addition. Each triangular block, as depicted in **Figure 11**, embodies a right-angle triangular shape, with dimensions provided in the diagram. The design of the triangular block is meticulously crafted to translate the horizontal motion generated by the linear actuator into vertical downward movement, facilitating the application of force to compress the flexible tube. The height of these triangular blocks dictates the extent of their downward motion, crucial for effective squeezing. Consequently, it is imperative to ensure that the downward movement of the blocks aligns with the diameter of the tube to achieve complete compression. Additionally, two rectangular slots adorn each block, facilitating smooth sliding along the rectangular guide affixed to the base. Positioned with their hypotenuses facing the linear actuator, the blocks measure a height of 1.8 cm, ensuring optimal compression of the flexible tube to propel materials efficiently.

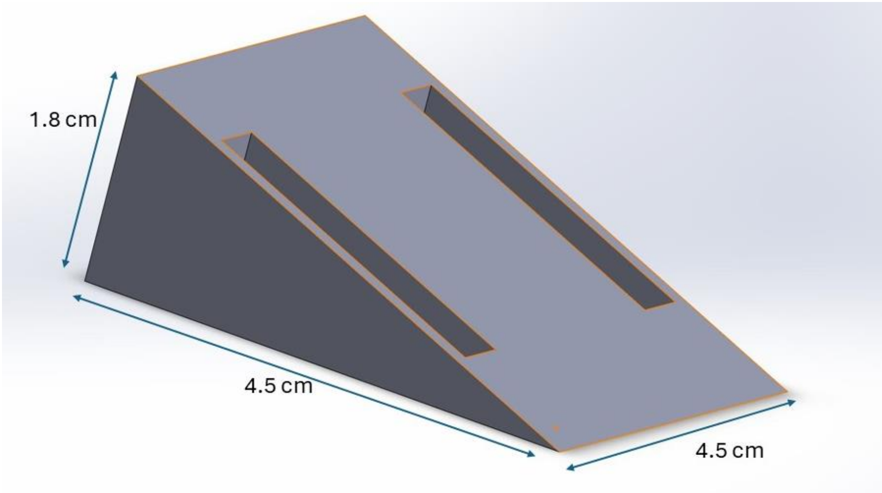

Figure 11: Triangular Block CAD Design with Dimensions

The base, illustrated in **Figure 12**, serves as the cornerstone of the mechanism that features a designated slot to house the linear actuator, with six rectangular holes strategically placed to secure the actuator using Velcro straps. Additionally, the base comprises four rectangular guides, each standing at a height of 6 cm, providing a pathway for the vertical movement of the triangular blocks. Positioned between these guides, the tube finds its placement within the mechanism. The base measures 8 cm in width and 30 cm in length.

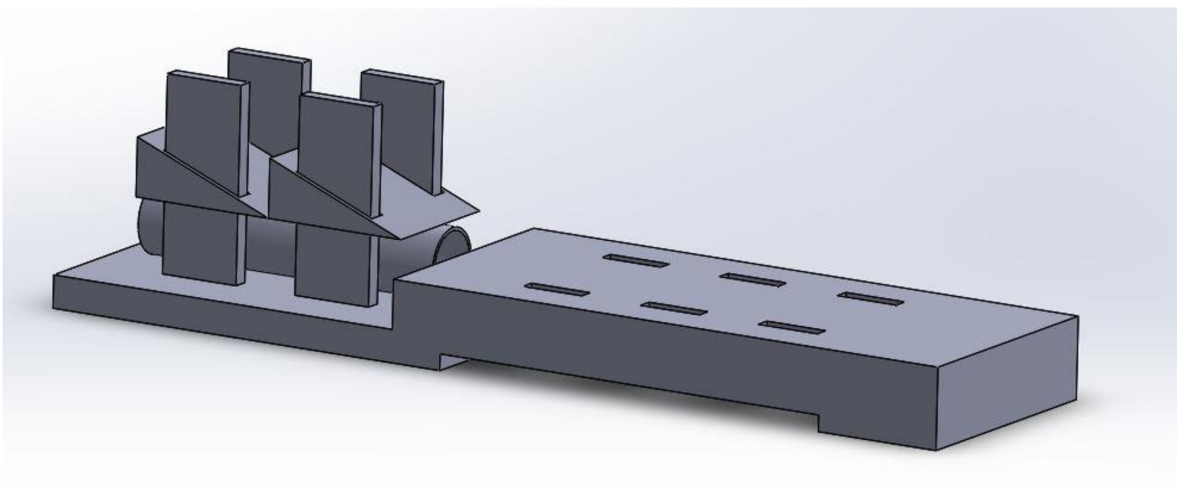

Figure 12: CAD Design of the Base with 2 Triangular Blocks and the Tube

Despite its effectiveness, the design presents certain risks and constraints. The sharp edges of both the base and triangular blocks pose a potential safety hazard, which could be mitigated by designing more rounded components in future iterations.

Additionally, determining the precise distance the blocks travel downward to exert pressure on the tube without causing damage is critical. Furthermore, the linear nature of the device restricts its application to straight tubes without curves due to the linear actuator's movement.

3.3.2. Triangular Block Peristaltic Mechanism Methodology

In crafting the Triangular Block Peristaltic Mechanism, a systematic methodology guided the design and fabrication process. Initially, Solidworks software was leveraged to meticulously engineer the intricate components of the mechanism. This powerful CAD tool provided the flexibility and precision required to bring the conceptual design to life, ensuring compatibility and functionality among the various elements.

Once the design phase was complete, Ideamaker software was employed to slice the intricate design into printable layers, optimizing the printing process for enhanced efficiency and accuracy. This slicing software enabled fine-tuning of parameters such as layer height and print speed, further refining the manufacturing process to achieve desired outcomes.

Subsequently, the Raise3D Pro 2 3D printer was enlisted to materialize the meticulously crafted design. Utilizing PLA filament with a diameter of 1.75 mm, the printer adeptly translated the digital blueprints into tangible components, layer by layer, with precision and reliability.

The printing process was methodically executed, with the base component requiring approximately 16 hours to complete its fabrication. In contrast, two of triangular blocks demanded approximately 2 hours and 20 minutes of printing time.

Lastly, the Velcro straps were purchased from a store to assemble the final prototype device, completing its functionality.

3.3.2.1 Linear Actuator

The selected linear actuator illustrated in **Figure 13** for the prototype device has the following specifications: it operates on DC 12 volts, featuring a stroke length of 150mm and a maximum travel speed of 10 mm/s. With a retracted length of 260mm and an extended length of 410mm, this actuator exhibits a maximum current draw of 3Amps and a maximum load-bearing capacity of 202lbs (900N). Crafted from high-quality aluminum alloy material, the actuator is both sturdy and aesthetically appealing, housing a high-performance motor with a built-in high-quality limit switch. Manufactured by Maopiner, the actuator operates with a low noise level below 50 dB and has a duty cycle of 10% [29].

This specific linear actuator aligns well with the application, providing ample force to facilitate the downward movement of triangular blocks. Its accessibility on Amazon, priced reasonably at $44.95, further adds to its suitability for the prototype. While the actuator's dimensions may be relatively large for a prosthetic application, the primary focus of this project is to validate the concept of a peristaltic pump for soft solid materials. The actuator's size becomes a secondary concern in this context, emphasizing the proof-of-concept nature of the prototype. In an ideal design for a prosthetic application, a more compact and miniaturized linear actuator with equivalent force capabilities would be preferred.

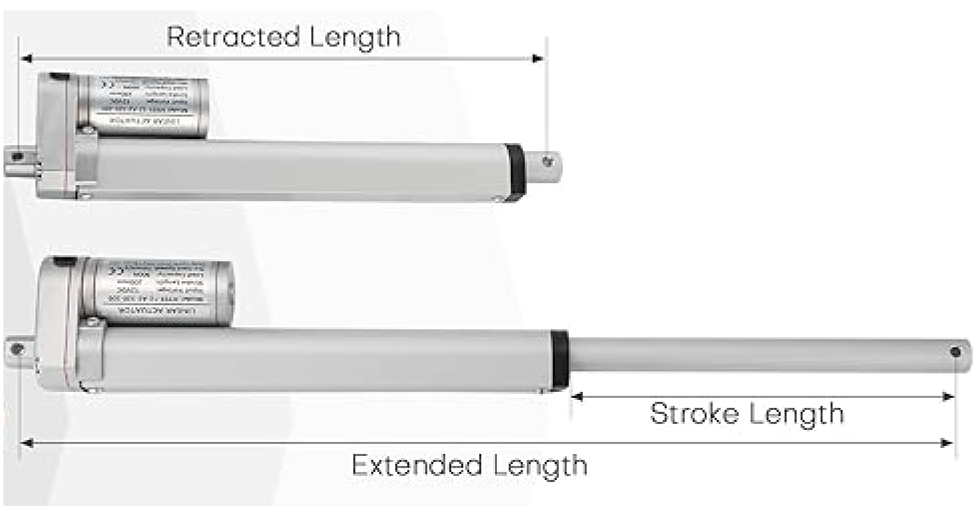

Figure 13: Selected Linear Actuator [29]

3.3.2.2 Electronics

In pursuit of simplicity, the electronic components employed in the prototype have been kept to a minimum. The primary electronic element is the linear actuator, as detailed in the preceding section. Accompanying this, a pivotal component is the switch, depicted in **Figure 14(a)**, responsible for controlling the extension and retraction of the linear actuator. The chosen switch functions as a relay, facilitating a change in polarity.

The switch's dimensions are visually presented in **Figure 14(b)**. Specifications of the switch are outlined as follows: DC 12V and 10A for voltage and current, double pole double throw (DPDT) for switch type, self-resetting for operation type, and 3 positions of ON-Off-ON. The wiring configuration involves connecting the red wire to the positive pole, the black wire to the negative pole of the power supply, and the blue wire to the motor. This setup enables the DC motor to reverse its direction through the switch, achieving polarity reversal. Simple connection of power, ground, and the two DC motor leads to the switch results in a reversal of the motor direction. The manufacturer of the switch is Dkardu, based in China [30].

Figure 14: (a)Switch Image - (b) Switch Dimensions [30]

The switch will be interfaced with a benchtop power supply adjusted to 12V with a current limit set at 3A for testing purposes. This configuration ensures controlled testing conditions and allows for a thorough examination of the switch and linear actuator’s functionality within specified voltage and current parameters.

3.3.3. Alternative Peristaltic Mechanism Designs

In the realm of alternative peristaltic mechanism designs, one prominent option mirrors the research study discussed earlier by Carpi et. al., featuring radially expanding flexible tubular modules constructed from DEs. While this design aligns with the peristalsis movement of the GI tract and has compact sizing suitable for implantation, concerns arise regarding its ability to exert adequate pressure on soft solid materials for efficient feces propulsion.

Before delving into the details of the four alternative peristaltic mechanisms, it's essential to explore diverse approaches in designing a peristaltic movement mechanism. The primary goal is to find an effective and reliable method for propelling soft solid materials, simulating the natural peristaltic movement, particularly in the context of a prosthetic colon application. Each design presents its unique features, advantages, and challenges, contributing to a comprehensive understanding of the alternatives available for consideration. Now, each mechanism will be briefly examined.

1. **Oblong Slider-Crank Mechanism:** This mechanism transforms circular motion into an oblong shape, leveraging a DC motor, crank, connecting rods, and a piston. While offering simplicity and reliability, its spatial requirements pose challenges for miniaturization, crucial for medical devices. Technical risks involve potential malfunctions and system halts, requiring further exploration. A visual representation of this mechanism is depicted in **Figure 15.**

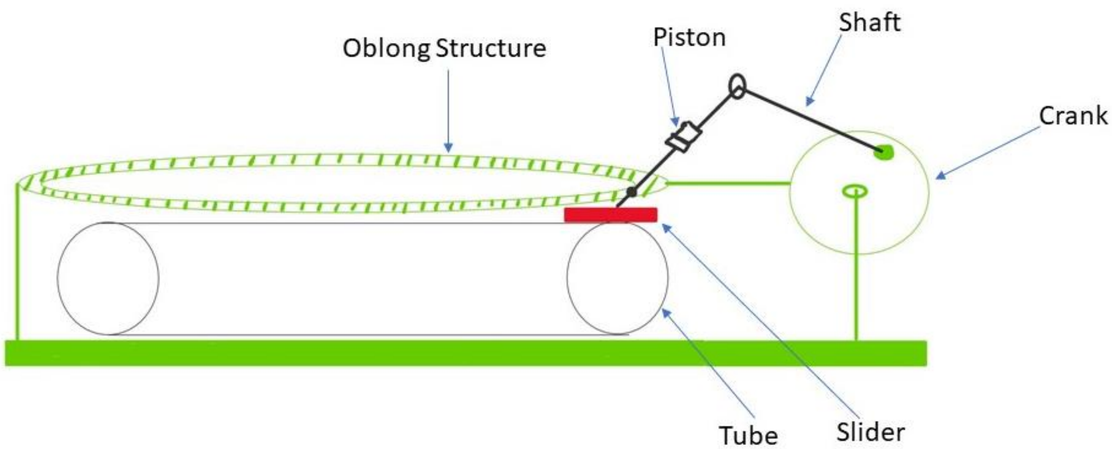

Figure 15: The Drawing of the Oblong Slider-Crank Mechanism

2. **Automated Ratchet Clamp Mechanism:** Employing a scissor-like clamp and a linear actuator, this design propels fecal matter forward by squeezing the tube. A ratchet mechanism facilitates unidirectional movement, and the system features simplicity and reliability. Risks involve potential challenges in transitioning between opening and closing positions and locating a robust spring for sufficient pressure. A visual depiction of this mechanism is presented in **Figure 16**.

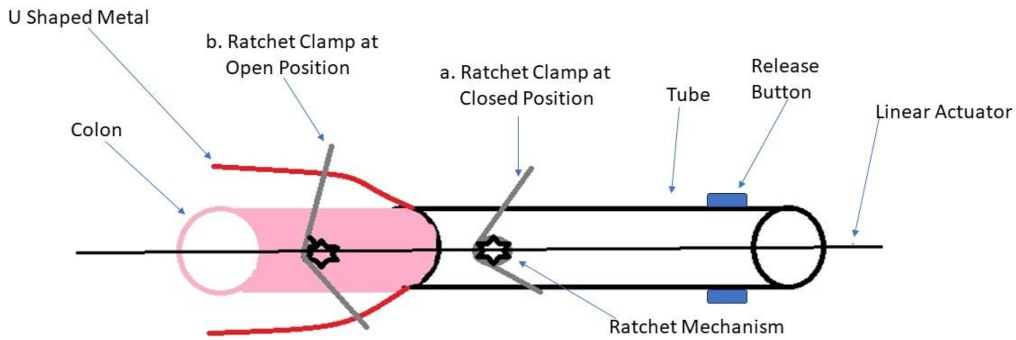

Figure 16:The Drawing of the Ratchet Clamp Mechanism

3. **Automated Mini Press Mechanism:** Utilizing a C-clamp mechanism automated with two linear actuators, this design provides precise control over pressure. The incorporation of two smooth surfaces reduces friction, but the addition of an extra motor increases system complexity and power consumption. In addition, press mechanism is slower in comparison to other design which can increase the risk of material blockage inside the tube. A visual representation of this mechanism is available in **Figure 17**.

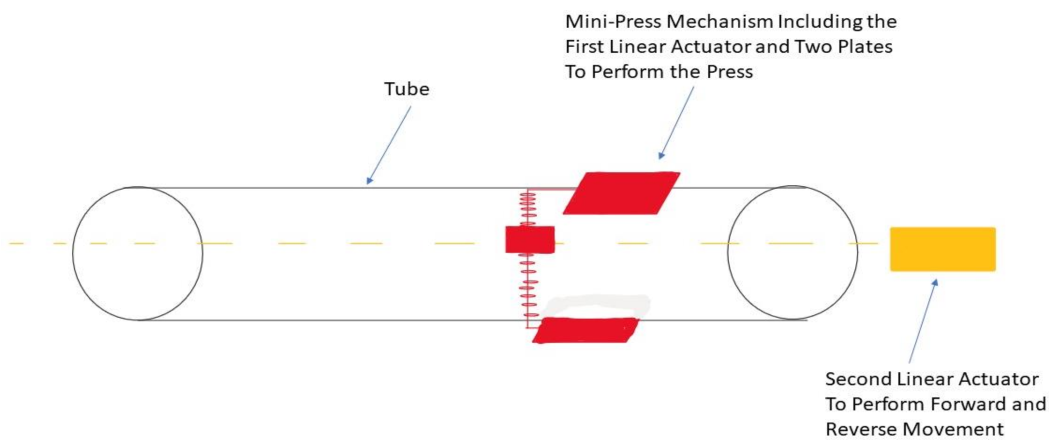

Figure 17:The Drawing of the Mini-press Mechanism

4. **Worm-Geared Clamp Mechanism:** Featuring an automated worm-geared clamp driven by a DC motor, this design ensures precision control over tube diameter. While excelling in compression, it introduces complexity, higher power consumption, and potential challenges in maneuvering the clamp due to friction.

The primary drawback shared among all the alternative designs mentioned is the necessity for the squeezing component of the device to slide back and forth along the tube. This arrangement introduces significant friction, which can lead to various risks, including malfunctions due to reaching the linear actuator's limit and potential damage to the tube. To mitigate these concerns, the decision was made to implement a triangular block mechanism that eliminates the need for sliding along the tube. This choice not only addresses the friction issue but also contributes to the simplicity of the design, minimizing the number of components to avoid unnecessary complexity.

Chapter 4. Prototype, Testing, and Discussion

Chapter 4 provides an in-depth examination of the prototype, elucidating the integration of its components. The discussion encompasses the final assembly of the device and explores viable options for fecal substitutes tailored for testing purposes. Additionally, this chapter delves into the intricacies of the testing setup, the methodology employed during the tests, and the systematic collection of data. The subsequent section critically analyzes the obtained results, offering valuable insights into the performance and functionality of the prototype.

4.1. Prototype

This section provides a comprehensive overview of a complete cycle of the device's operation, detailing each step in the process. Additionally, various substitution options for fecal matter will be explored and discussed within this context.

4.1.1. A Full Cycle of Device's Operation

The operational cycle of the device initiates with the linear actuator in full retraction mode. At this point, a feces substitute is positioned inside of the flexible tube. Upon pressing the activation button, the linear actuator extends its stroke, exerting force on the hypotenuse of the first triangular block. As the stroke progresses, the first triangular block descends, fully compressing the flexible tube beneath it.

As the first triangular block compresses the tube, the material inside undergoes propulsion in both directions. Upon complete descent of the first block, the stroke proceeds to the second triangular block, initiating force application on its hypotenuse and causing it to descend. At this point, any material previously propelled forward by the first block continues its forward motion exclusively. This occurs because the first triangular block fully compresses half of the tube, preventing material from traveling backward.

As the linear actuator reaches full extension, the reverse button is pressed to retract the stroke. The elasticity of the tube, prompt the triangular blocks to return to their relaxed positions, allowing the tube to expand to its natural diameter. This completes one full peristalsis movement cycle.

To continue the process, additional material is introduced into the tube, and the same sequence of steps is repeated.

4.1.2. Fecal Substitute

The next pivotal step involves testing the device with an appropriate fecal substitute that accurately mirrors the essential characteristics of human waste. To uphold safety and sanitation standards during testing and experimentation, it is crucial to employ materials that closely resemble real human feces. Various options exist for creating fecal substitutes that serve as reliable alternatives for experimental purposes.

One option involves preparing a mixture comprising water, oatmeal, and corn syrup. This blend offers a practical and cost-effective solution for crafting a fecal substitute. By adjusting the ratio of water in this mixture, the consistency can be tailored to simulate different scenarios effectively.

Another option entails preparing clay with customizable water addition, enabling the replication of various stages or conditions such as diarrhea or constipation. This clay-based mixture provides the versatility needed to accurately mimic different fecal consistencies. To compare the density of clay to fecal matter, it's essential to consider their respective densities. The density of clay can vary depending on its composition and moisture content but typically falls within the range of 1.6 to 2.6 grams per cubic centimeter (g/cm^3) [31]. On the other hand, the density of normal fecal matter ranges from about 0.9 to 1.2 grams per cubic centimeter (g/cm^3) when it is in a solid form [32]. Comparing these values, we observe that clay generally has a higher density than fecal matter. Despite this difference, clay remains a suitable substitute for fecal matter in experimental settings due to its malleability, ease of manipulation, and ability to replicate certain characteristics of feces, such as texture and consistency. Moreover, by adjusting the solid-to-liquid ratio during preparation, the density of the clay mixture can be further tailored to closely match that of fecal matter, ensuring accurate simulation in testing environments.

Once the chosen fecal substitute is prepared to match the desired properties, it can be poured into the flexible tube for comprehensive device testing. These simulated materials offer a hygienic and safe means to evaluate the device's functionality and ensure its efficacy in replicating peristaltic movement, obviating the need for actual human feces.

4.2. Test Setup and Procedure

The test setup for evaluating the prototype device is straightforward. The linear actuator switch is connected to a 12V bench power supply with a current limit of 3A. Following this, the prepared clay is inserted into the tube and then positioned beneath the triangular blocks. To initiate the test, the activation switch is pressed, causing the linear actuator to begin its extension. Once the linear actuator reaches full extension, the switch is toggled to retract the actuator back to its initial position.

To assess the device's performance, a proposed measurement methodology can be employed. This involves measuring the weight of the clay placed inside the tube and comparing it with the weight of the clay that exits the tube in a single stroke. For instance, if 100 g of clay is initially placed inside the tube and the output is measured to be 50 g, the device's performance can be calculated as 50%. Additionally, the time taken for one stroke can be measured to determine the device's flow rate. For example, if the flow rate is measured at 5 grams in 10 second, then the flow rate can be calculated at 0.5 grams per second which provides valuable insight into the device's efficiency.

4.3. Test Results and Discussion

In order to conduct the measurements, the flexible tube is first filled entirely with clay. The weight of the tube, including the clay, is then measured using a scale. To find the total weight of the clay in the tube, the weight of the empty tube (7.3 g) is subtracted from this measurement. This initial weight is recorded in **Table 2**.

Next, a stopwatch is used to measure the time it takes for the linear actuator's stroke to perform one full stroke, by a full extension and retraction. This stroke time is also recorded in **Table 2**.

Once the stroke is completed, the linear actuator is fully retracted to measure the output clay from the end of the tube. The exited clay is cut from the remaining clay inside the tube, measured on the scale, and recorded in **Table 2** under output weight.

The device's performance is then calculated using the formula $performance = \frac{Output\ Weight}{Initial\ Weight} \times 100$ and expressed as a percentage in **Table 2** under performance.

Finally, the flow rate is calculated using the formula $Flow\ Rate = \frac{Output\ Weight}{Stroke\ Time}$ expressed in grams/second and recorded under Flow Rate in **Table 2**.

A total of 8 trials were performed, and all the results were recorded in **Table 2**. Additionally, an example of the exited clay from one of the trials is depicted in **Figure 33**.

Trial	**Initial Weight ± 0.5 (g)**	**Output Weight ± 0.5 (g)**	**Performance (%)**	**Stroke Time (s)**	**Flow Rate (grams/second)**
1	57.3	4.50	7.85	12.0	0.38
2	57.2	5.50	9.61	12.10	0.46
3	61.5	7.20	11.70	11.60	0.62
4	54.2	3.60	6.64	10.16	0.35
5	61.2	4.00	6.54	12.05	0.33
6	61.1	6.5	10.64	13.00	0.50
7	60.9	3.2	5.25	11.72	0.27
8	60.9	3.1	5.09	11.95	0.25

Table 2: Test Results of Peristaltic Pump Prototype Trials

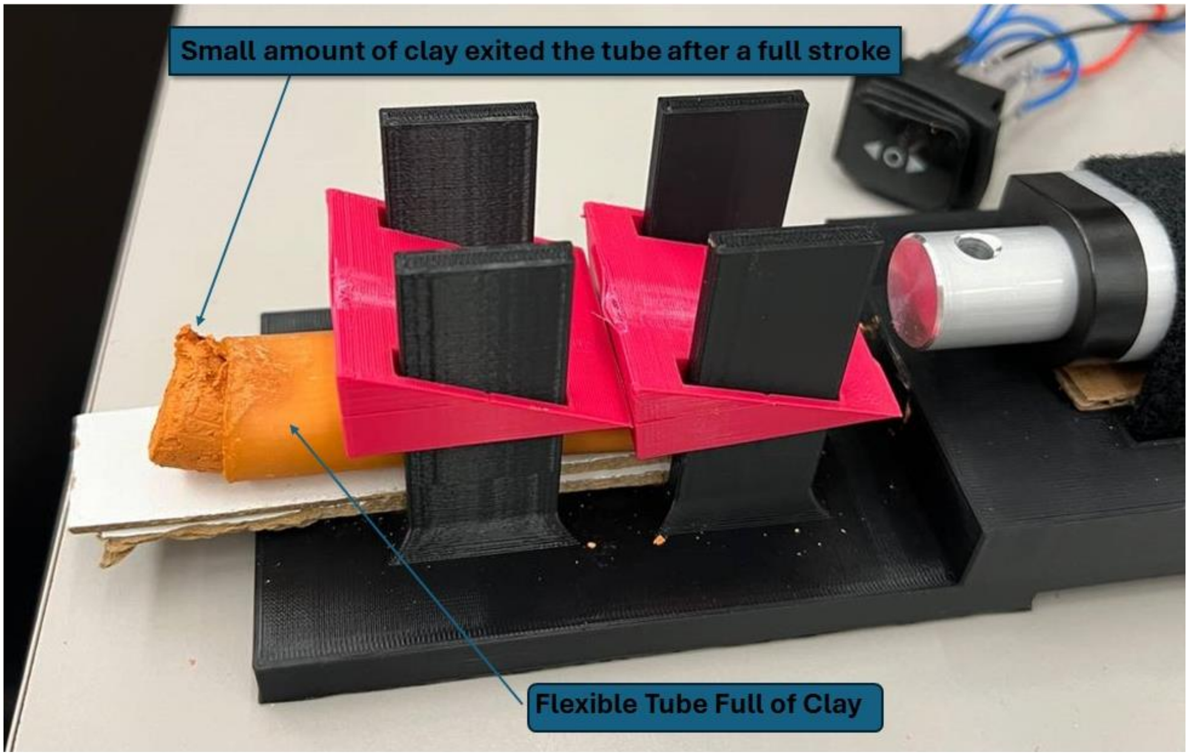

Figure 18: Example of exited clay shown on the left side after a full stroke in one of the trials

The performance expectations for the device are notably higher than the current results, driven by the objective of replicating the colon's natural function by efficiently expelling the majority of materials contained within the tube. The device is expected to expel approximately 48 grams of clay per stroke. This calculation is based on the volume of the tube compressed under the two triangular blocks. With 9 cm of the tube's height compressed by the two triangular blocks and a diameter of 2 cm, approximately 28 cm^3 of material is fully compressed, resulting in the propulsion of all the materials within that volume. Considering the density of the clay is approximately 1.7 g/cm^3, this equates to around 48 grams of clay being expelled per stroke. Therefore, the expected performance of the device is estimated at 80%. This anticipation was grounded in the device's primary function of transferring the majority of materials inside the tube from one end to the other. However, the current prototype's performance falls significantly below these expectations, ranging only between 5-12% in the conducted trials.

An observed dependency emerges regarding the output weight, notably influenced by the initial filling density of the flexible tube with clay. The trials indicate that a denser packing of clay within the tube correlates with a greater quantity of clay exiting after each stroke. This dependency on clay compaction poses a limitation as the device's efficacy becomes contingent on the uniformity of clay filling. Ideally, the device should exhibit consistent performance regardless of the initial clay distribution within the tube. **Figure 19** underscores this relationship, affirming a positive linear trend between tube compaction and output weight. The wide range of values surrounding the linear trendline in the graphs can be attributed to both the limited number of trials conducted and the narrow range of values observed for the initial weight (ranging from 54 to 62 grams) and output weight (ranging from 3 to 7.5 grams).

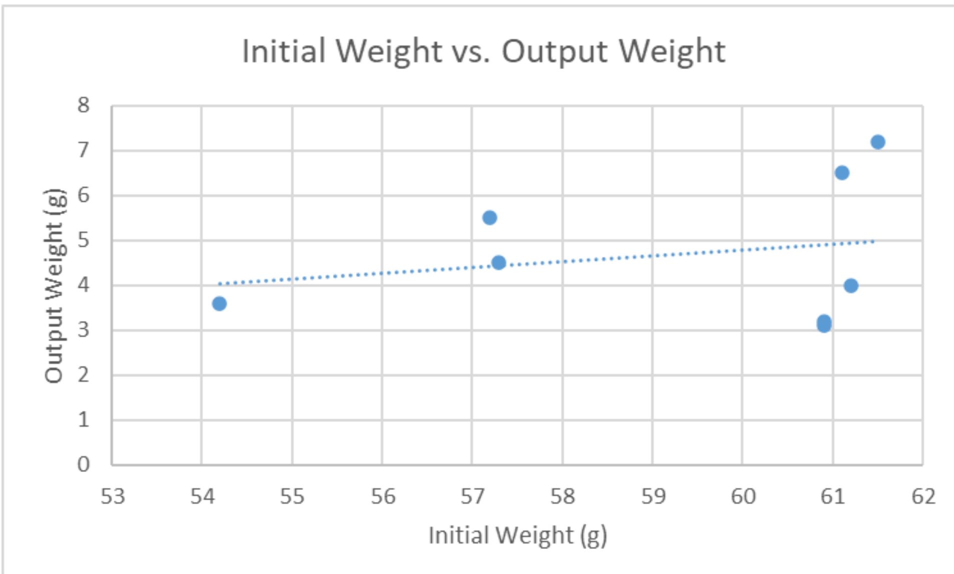

Figure 19: Positive Linear Behavior of Initial Weight vs. Output Weight

Despite the underperformance in output weight, the flow rate of material demonstrates reasonably consistent results. The flow rate is notably impacted by the stroke time, which in turn is influenced by the performance and speed of the linear actuator. Rose et al. estimated that the average adult produces approximately 128 grams of feces daily [33]. Given the current flow rate of the device, it would need to operate the peristaltic pump for approximately 5.4 minutes, which is quite reasonable. Furthermore, with the design adjustments discussed later in this section, the flow rate can be significantly improved, thereby reducing the activation time of the device.

Figure 20 (a) illustrates the flow rates across all trials and **(b)** charting the output weights over the course of the trials. The downward trend depicted by the negative linear trendline in both graphs indicates a decline in the device's performance over the course of the experiment. This phenomenon can be attributed to the occurrence of mechanical resonance in certain components of the device during testing, adversely impacting its functionality.

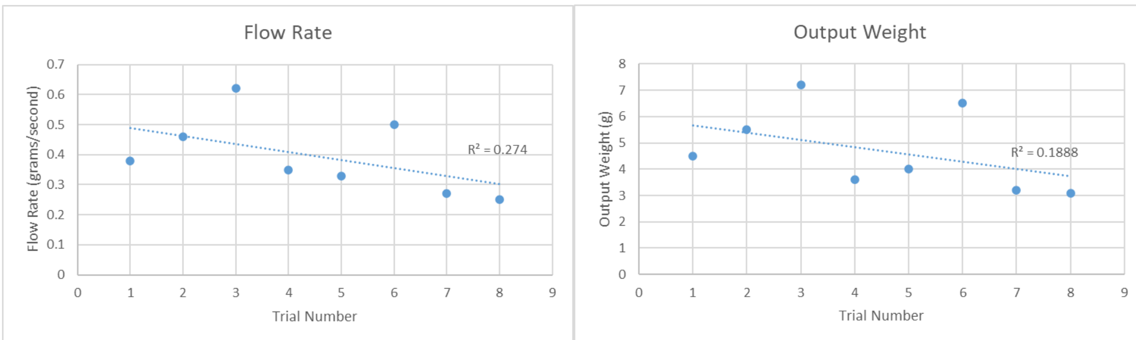

Figure 20: Graphs of (a) Flow Rate and (b) Output Weight over the Performed Trials

As previously highlighted, enhancing the device's performance to operate as a proficient peristaltic pump capable of expelling a more substantial amount of material is imperative. Throughout the experimentation phase, several design flaws and issues with the current prototype became apparent, necessitating adjustments for the subsequent prototype iterations.

One notable flaw observed is the displacement of the second triangular block when pressure is applied to the first block by the linear actuator. The displacement occurs because when the first triangular block compresses the tube, the clay propelled in forward direction causes the tube to expand, lifting the second triangular block. This misalignment impedes the smooth transition of the stroke, requiring manual adjustment to ensure continuous force application. Redesigning a threshold block mechanism on the guides could mitigate this issue by preventing upward movement of the triangular blocks beyond a specified threshold.

Furthermore, the lifting of the triangular blocks' tips at the end of each stroke results in uneven pressure distribution along the flexible tube, diminishing performance. This inconsistency is compounded by the large size of the blocks, which encounter heightened resistance during downward force application due to higher surface area in contact with the tube. Addressing these dimensional flaws by redesigning the blocks with smaller dimensions thus increasing the number of triangular blocks and their height could facilitate more uniform pressure distribution and enhanced material propulsion.

Strengthening the Velcro straps is essential to prevent mechanical resonance-induced lifting of the linear actuator, which disrupts the desired force application angle. Choosing more robust materials for securing the linear actuator to the base can mitigate this issue effectively.

Additionally, structural integrity concerns arose, notably the snapping of one triangular block's top during the final trial due to inadequate filling settings on the slicing software. Adjusting the printing settings to ensure thorough filling within the blocks will enhance their strength and reliability under the exerted force.

Moreover, the stiffness and stickiness of the clay present challenges to the peristalsis motion, hindering device performance. Opting for softer, less adhesive materials in future experiments can alleviate resistance, facilitating smoother material expulsion.

Incorporating these refinements in the next prototype iteration holds promise for optimizing device functionality and addressing the identified design limitations effectively.

Chapter 5. Challenges, Future Work, and Conclusion

In Chapter 5, challenges encountered throughout the design and development phases of the peristaltic pump for soft solid materials are thoroughly examined. The discussion extends to potential avenues for future enhancements aimed at refining the device's performance. The chapter concludes with a concise summary.

5.1. Challenges

One of the initial challenges encountered revolved around identifying a material that struck the delicate balance between flexibility and durability for the flexible tube application. This necessitated extensive experimentation with various materials before determining the most suitable option.

The quest for the optimal peristaltic design presented another significant challenge. While several designs were explored, each boasting its own set of advantages and drawbacks, ultimately, a single design had to be selected to propel the project forward.

Navigating the proof-of-concept phase for the peristaltic design tailored to soft solid materials posed its own set of challenges. It was imperative to develop a design that not only met the project's objectives but also aligned with the requirements for a prosthetic colon application. This required careful consideration of design criteria and constraints, balancing the need for complexity against the imperative of simplicity. Moreover, gaining a comprehensive understanding of the anatomy of the large intestine and the mechanics of peristalsis in the human body proved to be essential undertakings.

Lastly, sourcing an appropriate substitute for fecal matter to test the device presented its own set of hurdles. While clay was ultimately chosen as a stand-in for feces, it falls short in fully replicating the consistency and other characteristics of real fecal matter.

5.2. Future Work

In the realm of future developments, a pivotal focus lies on the miniaturization of the device, essential for its adaptation as an implantable medical device. This involves exploring more compact yet potent linear actuators, possibly drawing inspiration from Carpi et al.'s study to enhance radially expandable linear actuators. Simultaneously, efforts will be directed towards downsizing 3D-printed components, including the base and triangular blocks, for optimal implantation feasibility.

In tandem with the advancements, a critical stride in future development involves rendering the device fully automatic. This entails integrating sensory capabilities to detect the arrival of fecal matter, triggering the peristaltic pump for seamless material propulsion. Once the tube is emptied, the device will seamlessly transition into an idle state, enhancing the overall efficiency and autonomy of the prosthetic colon. This autonomous functionality aligns with the broader objective of refining the device into a self-sufficient and user-friendly implantable medical-grade solution.

Another critical avenue involves the quest for a robust, reliable, and implantable medical-grade tube, potentially elevating the entire device to be constructed from implantable medical-grade materials. Additionally, a comprehensive exploration of power consumption dynamics will be undertaken, probing whether the device can operate autonomously as an implantable unit or necessitates external power sources.

The overarching objective of these future endeavors is to refine the device into a more fitting prosthetic colon, aligning it closely with the standards and requirements of implantable medical-grade devices.

5.3. Conclusion

In conclusion, this thesis has documented the design, development, and testing of a device with peristaltic pump action specifically engineered for soft solid materials, with a focus on its potential application as a prosthetic colon. Through experimentation and iterative refinement, significant strides have been made in understanding the intricate mechanics of peristalsis and translating this knowledge into a functional prototype. The proof-of-concept prototype device presented herein demonstrates the feasibility of implementing peristaltic pump action on substances such as clay, showcasing an average material ejection of 4.7 grams in average 11.82 seconds, with an average flow rate of 0.39 grams per second and a calculated device performance of 7.9%. The challenges encountered along the way, from material selection to design optimization, have provided invaluable insights that have paved the way for future advancements.

Moving forward, the pursuit of miniaturization, automation, and the integration of implantable medical-grade materials stands as critical objectives in further enhancing the device's functionality and usability. By addressing these challenges and continuing to refine the device, we aim to develop a prosthetic colon that not only meets but exceeds the standards of implantable medical-grade devices. Ultimately, the culmination of this research represents a significant step forward in the quest to improve the quality of life for individuals in need of such medical interventions, underscoring the potential for innovation and collaboration in the field of biomedical engineering.

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Appendix A.

Chitubox Model Angle and Placement for Printing the F80 Resin Flexible Tube

To enhance the success rate of printing with soft resin, such as the F39/F39T/F69/F80 series resin, it is essential to follow specific steps in Chitubox software before slicing the model. One crucial step is to avoid placing the model in the middle of the build plate. This is because soft resin has high viscosity and can easily stretch and deform when separated from the FEP film during printing. Instead, it is recommended to position the model at a certain angle to prevent the formation of a vacuum sucker effect. For guidance on model placement, please refer to **Figure A.1(a)** and **(b)**.

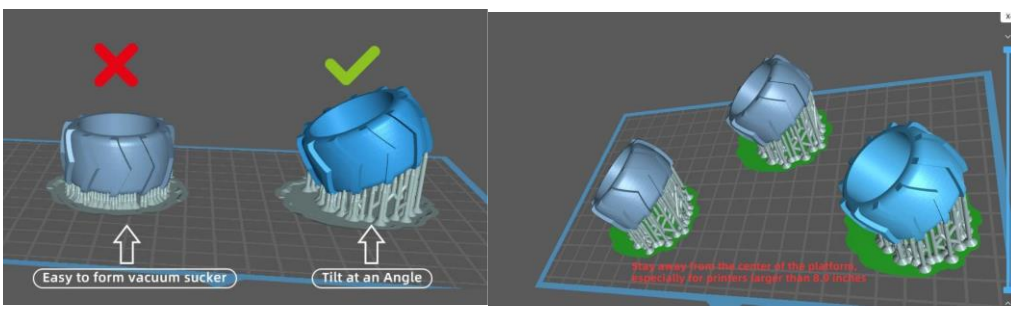

Figure A.1(a): The Angle Placement of the Model on the Build Plate (b) The Position of the Model on the Build Plate [22]

Appendix B.

Chitubox Support Setting for Printing the F80 Resin Flexible Tube

Utilize heavy and dense supports with a density ranging from 80% to 95%, oriented vertically (Angle: 0°). Ensure that the supports in the middle of the model are also thick. Refer to **Figure B.1** for the appropriate support settings for the model. Opt for a thin raft or no raft at all, as this facilitates easier removal of prints from the build plate, especially since F series resin tends to adhere tightly to the build plate surface.

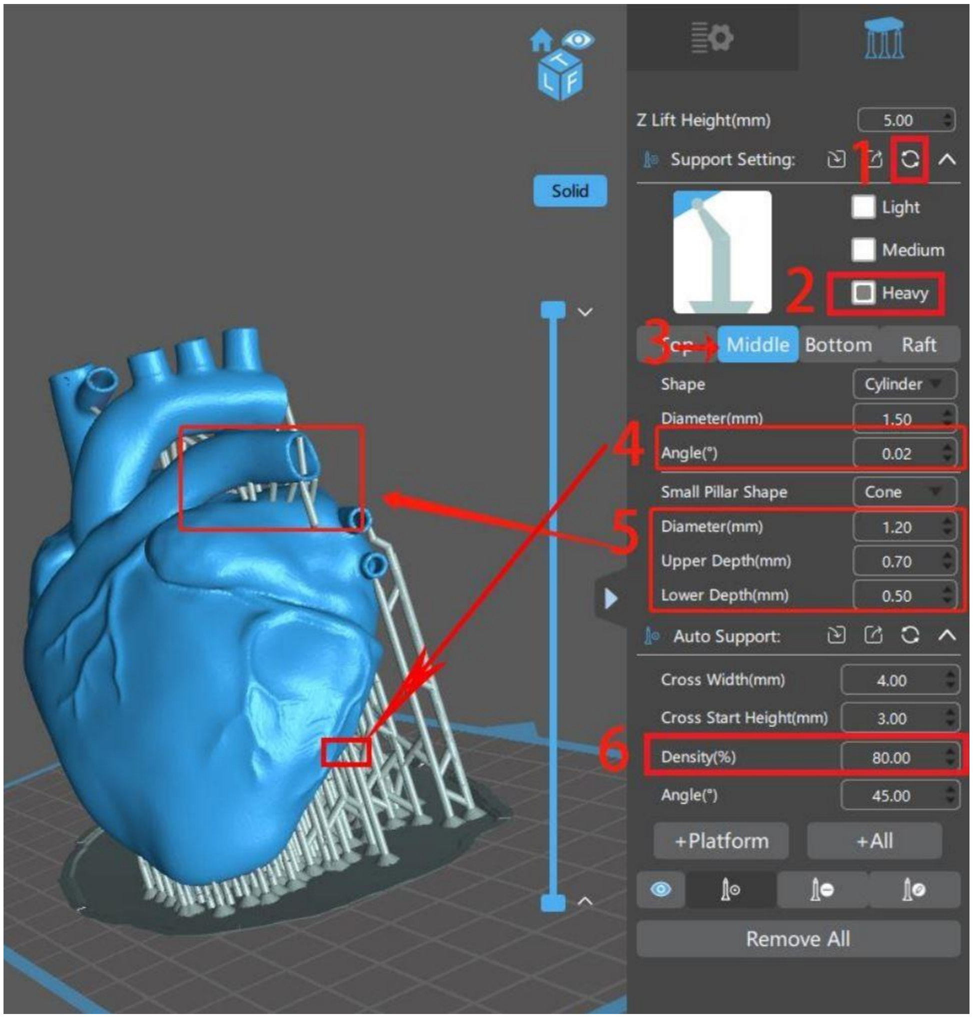

Figure B.1: Model Support Setting [22]

Appendix C.

Chitubox Printing Exposure Setting for Resin F80

To prevent stretching and deformation of the model, it is essential to employ a slow Lifting Speed. Additionally, a High Lifting Distance is necessary to ensure complete separation of the printing piece from the FEP film. A Long Light-off Delay time facilitates resin backflow. For optimal results, utilize the standard resin settings of the printer brand that has been successfully used before as the "Standard Settings" (25-35°C). If these settings are not available, revert to the default standard resin settings of the printer or slicer. Then, follow the method outlined below to adjust the settings of RESIONE resin according to the "Standard Settings" as detailed in **Table C.1** for successful printing.

	Standard Setting	**Resione F80 Resin**
Layer Heigh (mm)	50	50
Bottom Exposure Time (s)	A	1*A
Exposure Time (s)	B	1.2*B
Bottom Light-off Delay (s)	C	C+2
Light-off Delay (s)	D	D+2
Bottom Lift Distance (mm)	E	E+4
Lifting Distance (mm)	F	F+3
Bottom Lift Speed (mm/min)	G	0.5*G
Lifting Speed (mm/min)	H	0.5*H
Retract Speed (mm/min)	I	(2/3)*I

Table C.1: Printing Setting for Resione F80 [22]