

CONCEPTUAL DESIGN OF A TETHERLESS BISTABLE GRIPPING DEVICE FOR MINIMALLY INVASIVE PROCEDURES

by

Sai Pranav Avva

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Abstract

Minimally invasive surgery (MIS) has revolutionized modern clinical practice, reducing the trauma to patients, the time for recovery, and the operative risk. Most MIS tools are mechanically attached to the operator via rigid shafts and cables, limiting dexterity and access to tortuous anatomy and transmitting undesired forces to delicate tissue. To overcome this limitation, this thesis presents the conceptual design and prototype development of a tetherless magnetic gripping device for fluoroscopy-guided minimally invasive procedures.

The device proposed in this work is a bistable compliant shell with passive gripper prongs that decouple tissue engagement from continuous actuation by magnetic forces. Bistable snap-through enables rapid, low-energy transitions between open and closed states, so the device can grip and hold tissue without continuous energy input. Remote control of device orientation and state switching is achieved by magnetic actuation with an external coil system and embedded permanent magnets, eliminating the need for a mechanical tether.

Elastomeric and flexible resin materials were used to fabricate several design iterations to tune shell geometry, stiffness, and gripping performance. Experimental characterization was used to determine the actuation force

needed to induce bistable snap-through and to quantify the holding force attainable for representative phantom geometries. A complementary finite element analysis (FEA) study correlated shell thickness and curvature with snap-through force and stress distribution under loading, enabling more granular design optimization than experimental testing alone. The experimentally measured actuation requirements were compared to magnetic force estimates from the external actuation platform to identify key design tradeoffs and inform future geometric refinement.

The final prototype exhibited reliable bistable behavior, repeatable activation, and clinically relevant holding forces under simple phantom tests. However, the experimental and computational results presented here open up a pathway for further miniaturization, material optimization, and coil-platform integration, as the available magnetic force is not yet sufficient for fully autonomous actuation. Altogether, this work provides a basis for tetherless magnetic gripping devices with clinical applicability for tissue engagement, traction, and stabilization in minimally invasive surgical systems that could be operated under fluoroscopic guidance.

Thesis Committee

Primary Readers

Prof. Axel Krieger (Primary Advisor)
Associate Professor
Department of Mechanical Engineering
Johns Hopkins Whiting School of Engineering

Prof. Jeremy D. Brown (Academic Advisor)
Associate Professor
Department of Mechanical Engineering
Johns Hopkins Whiting School of Engineering

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Chapter 1

Introduction

1.1 Minimally Invasive Devices

The first laparoscopic appendectomy, widely regarded as a landmark in surgical history, was performed at the University of Kiel by the German gynecologist Kurt Semm [1], marking the beginning of the modern era of minimally invasive surgery (MIS). A review of over 37 PubMed-indexed articles from the years 2022–2025 noted that laparoscopy reduces average blood loss by approximately 30–60% and improves five-year survival rates in colorectal, gastric, and cervical cancers [2]. MIS has also facilitated access to surgery for higher-risk populations, such as elderly, obese, and chronically ill patients, significantly reducing postoperative pain and infection rates [2].

These advantages and the overall effectiveness of MIS arise not only from the surgical procedure itself, but also from the development of several specialized sub-technologies, including multifunctional tools such as articulating robotic wrist-like instruments with suction–irrigation probes. There have been significant advancements in miniaturized camera optics, 3D stereoscopic

camera systems, and near-infrared fluorescence imaging lenses [3]. Given that laparoscopy was first successfully performed in 1981 [1], substantial progress has been made in integrating CT/MRI navigation and heads-up display monitors. Energy-based devices, such as the Harmonic Scalpel and LigaSure, have further enhanced the precision and efficiency of MIS procedures [4].

The introduction of robotic teleoperation systems, such as the Da Vinci Platform, has enabled decoupling of the surgeon, tool, and patient proximities [5]. With accurate reproduction of the surgeon's actions at the tool end and near-real-time motion replication, there has been a shift in mindset from viewing surgical robots as passive assistance tools to regarding them as semi-autonomous, guided operative support systems [6].

Despite these advances, a key area for improvement is the mismatch between the human body's inherent softness and the rigidity of current tools. Most available instruments depend on physical tethers, namely wires, tubes, and rigid shafts. These not only restrict device dexterity but also limit anatomical accessibility, increasing the risk of tissue damage in highly deformable or tortuous regions of the body [7]. These limitations motivate research into developing tetherless, minimally invasive devices (MIDs) that operate without rigid physical connections [8].

1.2 Tetherless Systems and Actuation Strategies for Minimally Invasive Surgery

Richard Feynman famously articulated the goal of tetherless surgical systems in his seminal 1959 lecture at the annual meeting of the American Physical

Society, describing a thought experiment proposed by his colleague Albert R. Hibbs:

A friend of mine (Albert R. Hibbs) suggests a very interesting possibility for relatively small machines. He (Albert R. Hibbs) says that, although it is a very wild idea, it would be interesting in surgery if you could swallow the surgeon. You put the mechanical surgeon inside the blood vessel, and it goes into the heart and “looks” around. It finds the faulty valve and takes a little knife to slice it out.

[9]

This conversation hints at autonomous, self-contained devices that can function independently within the body. Here, *self-contained* refers to the integration of key subsystems—actuation, sensing, and communication—within the device itself. This design philosophy departs from traditional tethered systems, such as endoscopes, in which actuation mechanisms, imaging systems, and other mechanical components are positioned externally and connected to the surgical interface via rigid shafts or cables. Consequently, tethered platforms are subject to far fewer subsystem miniaturization constraints than tetherless devices.

In tetherless systems, device navigation, sensing, and data transmission must be performed wirelessly, which imposes stringent constraints on subsystem miniaturization. The reduced device scale enables navigation through sub-millimeter anatomical pathways while minimizing the risk of mechanical

tissue injury. In addition, these systems enable high-precision drug delivery and minimally invasive biopsy procedures, further reducing procedural invasiveness, recovery time, and postoperative complications. While these are compelling theoretical advantages, tetherless systems face practical engineering and manufacturing limitations. They must account not only for system scalability, but also for the fact that current manufacturing methods at the sub-millimeter scale are inherently based on 2D fabrication [9]. Devices operating at these scales must also remain biocompatible.

The actuation mechanism plays a critical role in determining the overall design characteristics and operational capabilities of tetherless devices. Consider, for instance, the commercially available PillCam capsule endoscope [10]: this tetherless device moves through the digestive tract using the body's peristaltic motion. As a result, it does not require an external power source for locomotion and uses onboard power primarily for data transmission, enabling a truly tetherless, miniaturized device. In general, the actuation methodology strongly constrains device size and target application. The following subsections review commonly investigated actuation modes for tetherless systems.

1.2.1 Magnetic Actuation

In magnetic actuation systems, device locomotion, orientation, and localization are governed by externally applied magnetic fields. To enable magnetic responsiveness, the device is either fabricated from magnetic materials or incorporates permanent magnets. This interaction between magnetic fields

and device architecture enables highly maneuverable tetherless systems with diverse operational capabilities. However, magnetic actuation also introduces important design constraints, particularly in material selection, as the device's mechanical components must minimize the use of magnetizable materials to avoid unintended magnetic interactions. The examples below highlight magnetically actuated tetherless devices developed for biomedical applications.

1. Rotating Permanent Magnets:

In this approach, an externally rotating permanent magnet generates a rotating magnetic field around the patient. Ye et al. [11] demonstrated that this rotating magnetic field can be converted into translational motion to propel a spiral-type capsule through the intestinal tract.

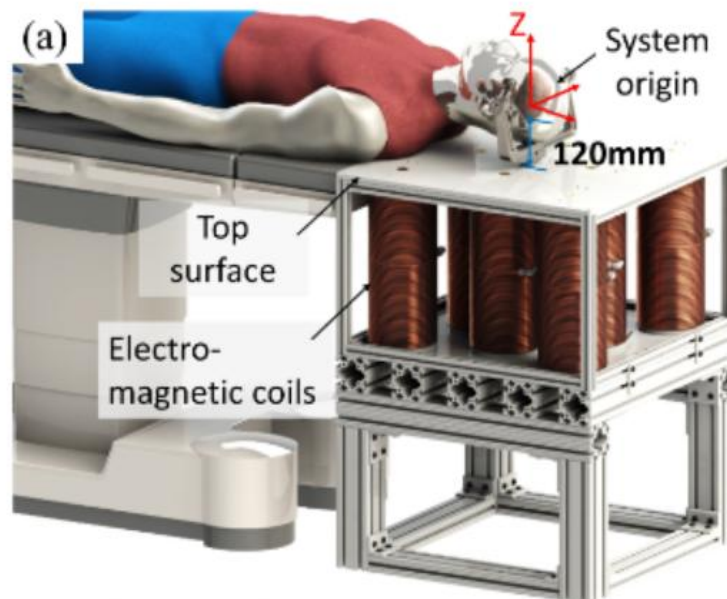


Figure 1.1: Magnetic capsule actuated by rotating magnets navigating through the intestine. Image Source: Ye et al. [11].

2. Electromagnetic Coil Arrays:

These systems utilize a series of independently controlled electromagnetic coils to generate a steerable magnetic field within a three-dimensional workspace, enabling navigation of tetherless magnetic tools. Schonewille et al. [12] developed a system comprising eight electromagnets positioned entirely beneath the operating table, thereby maintaining an unobstructed surgical workspace while enabling the deployment and steering of millimeter-scale magnetic microsurgical tools for deep-brain neurosurgical applications. The MagnetoSuture platform [13] demonstrated tetherless manipulation of magnetized suture needles via an external electromagnetic actuation system, including remote translation, orientation, and tissue penetration. A later platform, VORIACS [14], increased the available magnetic gradient and field articulation capabilities by using twelve independently controlled coils arranged into four concentric coil stacks.

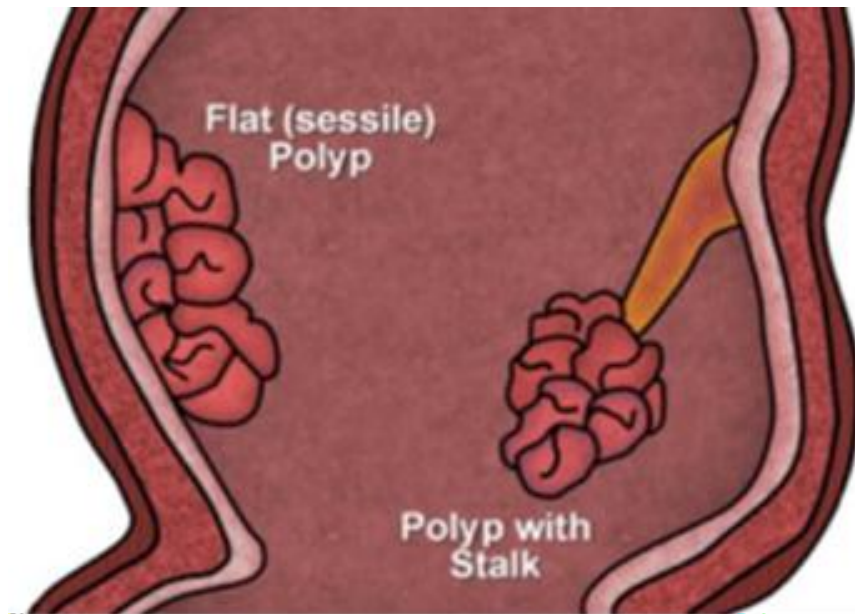


Figure 1.2: Electromagnetic coil array for tetherless device actuation. Image Source: Schonewille et al. [12].

3. MRI-Powered Devices:

MRI-powered devices utilize the gradient coils of an MRI scanner as both the actuation and localization source, eliminating the need for additional external hardware. Pouponneau et al. [15] employed this approach to develop therapeutic magnetic microcarriers for targeted drug delivery within the vascular system. The MRI platform enabled simultaneous device actuation and localization during operation.

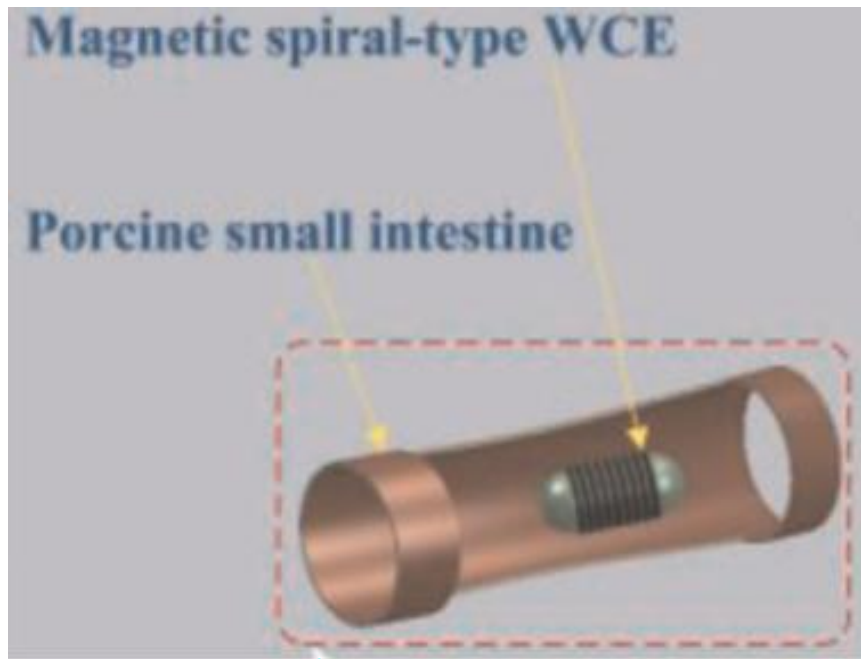


Figure 1.3: Therapeutic magnetic microcarriers. Image Source: Pouponneau et al. [15].

1.2.2 Thermal and Chemical Actuation

Thermal and chemical actuation methods for tetherless devices are generally passive, relying on physiological stimuli present in the body. These systems are designed to respond selectively to localized conditions at the target site, including temperature variations [16], pH [17], or other biochemical signals. Although thermal actuation may be externally assisted, its operational range is constrained by the need to minimize collateral thermal damage to surrounding tissues.

1. Thermally Triggered Actuation:

Dunn et al. [16] investigated thermomagnetic self-folding microgrippers for minimally invasive biopsy applications. The proposed device

was validated *in vitro* for fibroblast cell extraction. The system utilizes a thermoresponsive polymer layer that undergoes a reversible phase transition near body temperature, causing the microgripper's finger-like appendages to self-fold and encapsulate the target cells.

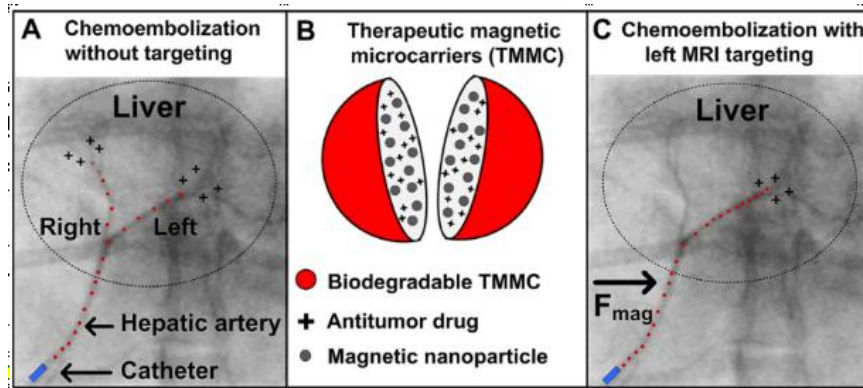


Figure 1.4: Thermally activated gripper for biopsy. Image Source: Dunn et al. [16].

2. Biochemically Triggered Actuation:

Biochemical actuation mechanisms exploit localized variations in physiological pH to trigger device operation. Hou et al. [17] investigated a targeted drug-delivery system activated by the mildly acidic tumor microenvironment (pH 5.6), in contrast to the physiological pH of approximately 7.4. The proposed device utilized pH-responsive nanomaterials coated with hydrophilic layers to enable controlled drug release at the target site.

1.2.3 Solvent and Hydrogel-Driven Actuation

Solvent- and hydrogel-driven actuation methods rely on passive, self-powered mechanisms that convert chemical potential energy, surface-tension differences, solvent miscibility, or local ionic/pH environment into mechanical motion. Their reliance on environmental changes makes them attractive for fully autonomous, tetherless microdevices operating within fluid-filled anatomical spaces. However, their actuation range and force output are typically limited by diffusion kinetics and the magnitude of the local chemical gradient.

1. Solvent-Driven (Marangoni) Motion:

Bassik et al. [18] demonstrated that lithographically patterned poly(N-isopropylacrylamide) (PNIPAm) hydrogels, when soaked in ethanol and transferred into water, generate spontaneous self-propelled locomotion driven by the expulsion and interfacial spreading of ethanol at the air–water boundary. This Marangoni-effect motion produced linear velocities up to 31 cm/s and rotational speeds up to 3529 rpm, without requiring onboard wiring, batteries, or external field control.

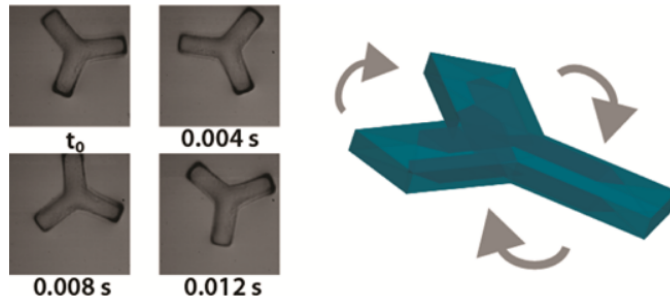


Figure 1.5: Marangoni-effect-driven locomotion of lithographically patterned PNIPAm hydrogels, showing translation, rotation, and precession modes generated by ethanol spreading at the air–water interface. Image Source: Bassik et al. [18].

2. Directional and Pathfinding Motion via Liquid Mixing:

Wang et al. [19] showed that submerged polymer hydrogels can be directionally propelled by spreading a miscible, low-surface-tension liquid across a water surface, with motion governed by convective flows arising from the resulting surface-tension gradient. Direction and velocity were tunable via the liquid-mixing conditions, and a spherical hydrogel exhibited autonomous pathfinding behavior, navigating a microfluidic maze toward the liquid-mixing exit without electronic or magnetic guidance.

3. Multiphoton-Fabricated Chemically Responsive Protein Hydrogels:

Kaehr and Shear [20] used scanning multiphoton excitation to photocrosslink proteins (BSA, avidin, lysozyme) into submicrometre-precision 3D hydrogel microstructures with tunable volume changes in response to pH, ionic strength, or specific ligand binding. High-aspect-ratio geometries such as rods and tethers achieved actuation response times under one second, and spatially graded crosslink density enabled programmed asymmetric bending, allowing designed folding and unfolding motions

analogous to those of a compliant bistable shell, but triggered entirely by chemical stimuli rather than by mechanical snap-through.

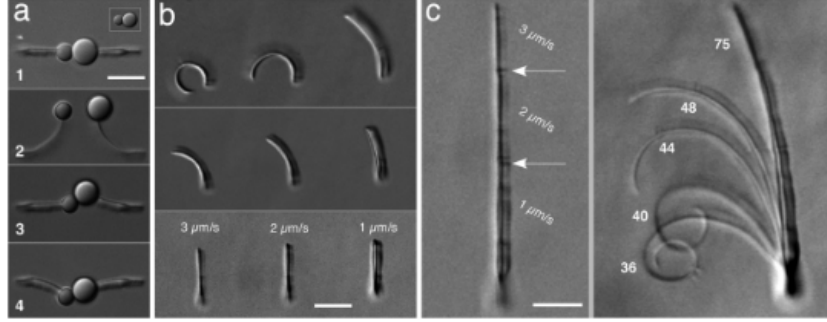


Figure 1.6: Multiphoton-fabricated protein hydrogel microstructures exhibiting chemically triggered folding and volume-change actuation. Image Source: Kaehr and Shear [20].

Collectively, these solvent- and hydrogel-driven mechanisms represent a distinct actuation category characterized by passive, battery-free, field-free operation. This offers a useful design contrast to the bistable magnetic gripper developed in this thesis: whereas solvent/hydrogel systems rely on continuous chemical gradients or one-time diffusion-limited triggers, the bistable snap-through mechanism provides a discrete, repeatable, force-efficient transition that can be actuated on demand via an external magnetic field, at the cost of requiring that field to be present and sufficiently strong at the target depth.

1.2.4 Acoustic Actuation

Focused ultrasound actuation enables transcutaneous transmission of high-frequency acoustic energy through biological tissue, allowing tetherless devices to be actuated at depth without a direct physical power connection. An additional advantage of ultrasound-based systems is their capability to

provide simultaneous localization. Hao et al. [21] demonstrated a focused ultrasound-actuated tetherless device capable of generating a peak output force of approximately 5.5 N, sufficient for tissue penetration and liquid cargo release. The system utilized an ultrasound-induced phase-transition mechanism to achieve controlled device actuation.

1.3 Gripper Devices

One of the fundamental tasks of untethered devices is interacting with the target site, a capability that this thesis addresses directly through its gripping mechanism. Such interaction is most commonly achieved through physical contact, which requires the device to grip or anchor itself to the target tissue. This gripping and anchoring capability forms the core functional requirement of the tetherless magnetic device developed in this thesis. A variety of gripper actuation strategies have been explored in the literature to achieve this, ranging from thermally and chemically responsive mechanisms to pneumatically driven and magnetically actuated compliant structures. The following subsections review these gripper actuation modalities, with particular attention to bistable and compliant designs relevant to the tetherless magnetic gripping device developed here.

1.3.1 Thermally and Chemically Triggered Grippers

Thermally and chemically triggered grippers are among the earliest and most extensively studied classes of untethered gripping mechanisms. Leong et al. [22] demonstrated tetherless, mass-producible microgrippers fabricated

from rigid polymer panels connected by a thermobiochemically responsive hinge layer, which can be remotely triggered to curl and close around a target object under physiologically relevant temperature or chemical stimuli. Because actuation is driven entirely by the local environment rather than an external field or onboard power source, these devices achieve full tetherless operation, though at the cost of relying on passive, relatively slow, and often irreversible triggering conditions.

Building on this thermoresponsive actuation principle, Gultepe et al. [23] developed untethered microgrippers capable of performing *in vivo* biopsy retrieval and validated the approach in a porcine bile duct model. The grippers were deployed into the biliary tract, triggered to close around target tissue upon reaching body temperature, and subsequently retrieved for histological analysis, demonstrating that thermally actuated grasping can achieve clinically meaningful tissue capture without any physical tether. However, both approaches exhibit gripping speeds on the order of seconds to minutes and offer limited control over grip timing or reversibility once deployed, motivating the exploration of more actively controllable actuation strategies for gripping tasks that demand precise temporal or spatial coordination.

1.3.2 Magnetically Actuated Grippers

Magnetic actuation addresses several limitations inherent in thermally and chemically triggered systems by enabling real-time, externally controlled gripping. Diller and Sitti [24] reviewed a broad class of stimuli-responsive soft untethered grippers for drug delivery and robotic surgery, highlighting

magnetic actuation as a particularly attractive strategy because it allows the surgeon or operator to trigger, hold, and release the gripping state on demand, in contrast to the largely irreversible or slow-acting thermal and chemical triggers discussed previously. Magnetically responsive grippers typically integrate soft or compliant structures embedded with magnetic particles or discrete permanent magnets, allowing an externally applied field to induce a controlled shape change that closes the gripper around the target tissue.

This actuation flexibility comes with its own design trade-off: most magnetically driven grippers require continuous application of the external field to sustain the closed, gripping configuration, since removing the field allows the compliant structure to relax back toward its original shape. This constraint motivates mechanisms that can retain a gripping state independent of continuous field application, a gap addressed by bistable structural designs.

1.3.3 Bistable and Multistable Grippers

Bistable and multistable mechanisms offer a structural solution to the actuation–retention trade-off observed in magnetically and pneumatically driven grippers. Chi et al. [25] provided a comprehensive review of bistable and multistable actuators for soft robotics, describing how snap-through behavior in constrained beams, curved shells, and compliant mechanisms allows a structure to transition between two (or more) mechanically stable configurations using only a brief input of energy, after which the structure remains in its new state without requiring continuous actuation. This property is particularly advantageous for gripping applications, since a bistable gripper can snap

into a closed, tissue-engaging configuration and remain stably closed even after the actuation input—whether magnetic, pneumatic, or mechanical—is removed.

The review further categorizes the range of soft actuating materials that have been paired with bistable elements, including magnetically responsive polymers, and notes their use in fast-grasping applications where rapid, energy-efficient state transitions are required [25]. This combination of bistability with magnetic responsiveness is directly relevant to the tetherless magnetic gripping device developed in this thesis, which leverages a bistable compliant shell to decouple the energy required to trigger gripping from the energy required to sustain it, thereby addressing a key limitation shared by both the thermally and chemically triggered grippers in Section 1.3.1 and the continuously actuated magnetic grippers in Section 1.3.2.

The development of a generalized gripper platform enables the specific device to be used across multiple procedures. In the following section, we consider a clinical application and how the gripper platform could be adapted for a variety of uses.

1.4 Clinical Motivation: Gripping Device for Colonic Polyps

1.4.1 Colonic Polyps and Clinical Significance

Colonic polyps are abnormal tissue growths that arise from the mucosal lining of the colon and rectum. Morphologically, polyps are commonly classified as either pedunculated or sessile, as illustrated in Figure 1.7. Pedunculated

polyps possess a distinct stalk connecting the lesion to the mucosal surface, whereas sessile polyps exhibit a broad base directly attached to the colonic wall. Sessile lesions are considerably more difficult to remove completely because they lack a well-defined attachment point for conventional snare-based tools. Histologically, adenomatous and serrated polyps are clinically significant as major precursors to colorectal cancer.

The prevalence and downstream consequences of colonic polyps make their complete removal one of the most important interventions in preventive gastroenterology. Large colonoscopy screening studies report adenoma detection rates of approximately 26–33% in screening populations [26]. Furthermore, the National Polyp Study demonstrated that colonoscopic polypectomy reduces colorectal cancer incidence by 76–90% and lowers colorectal cancer mortality by approximately 53% [27]. These findings establish complete polyp resection as a clinically critical procedure directly linked to colorectal cancer prevention and patient survival.

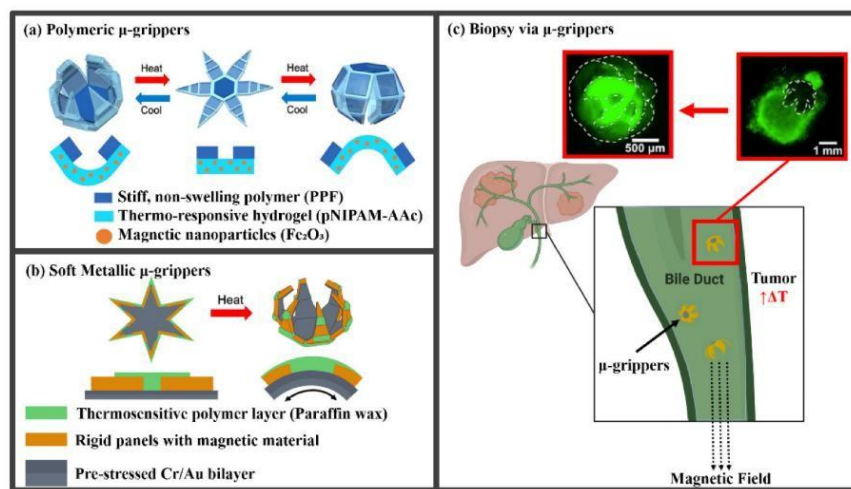


Figure 1.7: Types of tumors in the rectum. Image Source: Zauber et al. [27].

1.4.2 Limitations of Current Tethered Polypectomy Tools

Conventional polypectomy procedures are performed using tethered instruments, such as cold and hot snares, endoscopic mucosal resection (EMR), and endoscopic submucosal dissection (ESD), introduced through the working channel of a flexible endoscope [28]. Although effective in routine cases, these instruments are constrained by shaft stiffness, friction within the endoscope channel, and limited articulation capability [29]. In anatomically challenging regions of the colon, maintaining optimal cutting orientation relative to the lesion becomes difficult, particularly for sessile or flat polyps. As a result, incomplete resection remains a significant clinical challenge.

Previous studies report colorectal polyp miss rates of 17–28% [30], with anatomically difficult lesions and inadequate snare orientation as major contributing factors. Incomplete resection is also a primary contributor to interval colorectal cancer development [31]. In addition, complications such as post-polypectomy bleeding and perforation increase substantially during technically complex resections. These failure modes arise fundamentally from the rigid mechanical coupling between the cutting instrument and the external operator through a tethered shaft, which restricts maneuverability and transmits unintended forces directly to the surgical site.

1.4.3 Clinical Gaps Addressable by Tetherless Magnetic Devices

The limitations of conventional tethered tools define a clinical gap that may be addressed through tetherless magnetic surgical devices. Unlike tethered

systems, a tetherless device is not constrained by shaft orientation or friction within the endoscope channel, allowing it to access lesions in geometrically challenging regions of the colon. This is particularly advantageous for sessile polyps positioned behind folds or within sharply curved anatomical segments where conventional snares struggle to achieve proper orientation.

A tetherless magnetic gripping device additionally enables gripping and traction forces to be localized directly at the lesion site rather than transmitted through a long flexible shaft. In the proposed design, tissue engagement and stabilization are achieved through bistable gripping, allowing the device to anchor to the polyp or surrounding mucosa. At the same time, conventional snares or other resection tools are used to perform polypectomy. Such a mechanism may provide more uniform traction around the lesion base compared with conventional snare-based approaches, particularly for irregular or broadly attached sessile lesions. Localizing the gripping forces within the device may also reduce unintended traction on the colonic wall, thereby lowering the risk of perforation during complex resections.

Magnetic actuation provides several advantages for enabling tetherless operation in the colonic environment. Magnetic fields penetrate biological tissue with negligible attenuation and do not require a direct line of sight between the external actuation source and the implanted device. Consequently, magnetic systems enable wireless device actuation without onboard power-delivery wires or tethered energy transmission pathways. Furthermore, unlike thermal or chemical cutting modalities, magnetic actuation enables mechanical tissue interaction without depositing significant heat or reactive agents at the

surgical site, thereby reducing collateral tissue damage.

Previous work has demonstrated that magnetic actuation can support several minimally invasive medical tasks beyond simple locomotion. The MagnetoSuture platform established tetherless manipulation of suture needles, and subsequent work demonstrated semi-autonomous 2.5D needle control for suturing on tissue-mimicking phantoms [13, 32]. For gastrointestinal applications, Yan et al. [33] reported a mobile four-electromagnet platform mounted on a collaborative robot that used deep reinforcement learning to guide a 7 mm magnetic capsule over a clinically relevant planar workspace in a gastrointestinal phantom. In a cardiovascular application, Bhattacharjee et al. [34] demonstrated vision-guided autonomous control that maintained millimeter-scale trajectory tracking of a drug-carrying magnetic millirobot under pulsatile flow in an *in vitro* heart phantom. Together, these studies show that magnetic actuation can enable suturing, gastrointestinal navigation, and localized therapeutic delivery, motivating its use for a tetherless gripper intended for tissue engagement and stabilization.

1.4.4 Thesis Motivation and Proposed Design Direction

Despite recent advances in tetherless magnetic systems, a major limitation remains the relatively low mechanical output force achievable through externally applied magnetic fields. Onaizah and Diller [35] demonstrated magnetically actuated, tetherless microsurgical scissors; however, the generated forces were directly dependent on the applied magnetic flux density, and precise three-dimensional orientation control increased actuation complexity. These

constraints similarly limit the gripping force and reliability achievable with purely compliant, continuously actuated magnetic grippers.

In this thesis, the concept of tetherless magnetic actuation is extended to develop a bistable, tetherless *gripping* device for minimally invasive surgical procedures. The proposed work draws inspiration from previously reported magnetically actuated suturing systems and tetherless microsurgical scissors, while reorienting the design objective toward secure tissue engagement and traction rather than direct cutting. The primary design objective is to decouple tissue gripping from continuous high-magnitude magnetic actuation by combining a bistable compliant shell with passive gripping prongs. In this approach, magnetic actuation triggers a bistable snap-through. It establishes a high-force gripping state, after which the device can maintain tissue engagement with minimal or no additional actuation. This strategy enables gripping performance to be governed primarily by the bistable shell geometry and compliant interface rather than by the instantaneous magnetic field magnitude, thereby addressing one of the principal limitations of existing tetherless magnetic gripping devices.

Chapter 2

Conceptual Design

Chapter 1 established the clinical motivation for a tetherless magnetic gripper in colonic polypectomy, highlighting the limitations of rigid, tethered instruments and motivating a design that can securely engage tissue while operating within the geometric and force constraints of the colon. Building on this motivation, this thesis proposes a generalized tetherless gripping platform defined by three core design constraints:

1. **Three-dimensional gripping capability:** The device must grip in full 3D space rather than along a single fixed plane, enabling stable engagement with irregularly shaped, variably oriented polyps and surrounding mucosa within the colonic lumen.
2. **Bistable shell actuation:** Gripper actuation should be realized through a compliant bistable shell, so that transitions between open and closed states require low input force and do not depend on continuous high-magnitude magnetic actuation.
3. **High grip strength with low actuation force:** The platform must deliver

clinically relevant gripping forces (approximately 2–5 N for polyp stabilization) while keeping the required actuation force within the limits of external electromagnetic coil systems (on the order of 1.5 N) [13, 36, 37].

These constraints motivate the use of compliant, bistable mechanisms for passive gripping and directly inform the conceptual architecture developed in this Chapter.

2.1 Passive Gripping via Compliant Bistable Mechanisms

One of the primary technical challenges arising from the above constraints is achieving stable, repeatable grasping of irregularly shaped, variably stiff, and often slippery biological structures using a fully tetherless device. Conventional approaches to gripping typically rely on actively controlled mechanisms driven by motors, pneumatics, or cable transmission. In magnetically actuated tetherless devices, however, the incorporation of onboard motors or conventional actuation mechanisms is severely constrained by device size, power availability, and material compatibility with strong external magnetic fields. Consequently, achieving fully active gripping in a truly tetherless, miniaturized system is impractical.

To address this limitation, compliant passive gripping mechanisms present a promising alternative. Zhang, Sun, and Zhao [38] proposed a 3D-printed compliant bistable gripper for aerial perching and grasping, as shown in Figure 2.1. Fundamentally, the design philosophy of this system relies on structural compliance, in which flexible silicone tubes serve as torsional joints

in place of conventional rigid hinges. This approach demonstrates how elastic deformation within a structure can replace complex mechanical assemblies and enable passive grasping without conventional actuators.

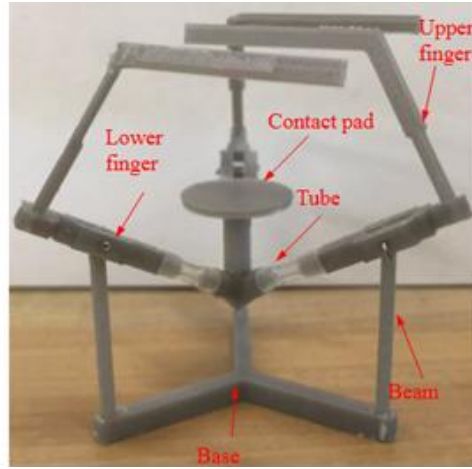


Figure 2.1: Compliant aerial gripper based on a bistable mechanism [38].

The bistable gripper proposed by Zhang, Sun, and Zhao [38] relies on bistability to transition between open and closed states through impact forces generated during contact. This eliminates the need for onboard sensors or active actuators. However, the mechanism relies on an aerial platform to provide the kinetic energy required to close. When adapted for a tetherless minimally invasive device operating within the body, such impact-driven state switching becomes difficult to realize, and an alternative triggering modality is required. Furthermore, this three-finger symmetric gripper configuration was experimentally validated primarily on cylindrical and geometrically constrained objects, thereby reducing its suitability for interaction with irregularly shaped biological tissues.

To address the limitations of impact-driven bistable grippers, inspiration can be drawn from the work of Balakuntala, Palathingal, and Ananthasuresh [39], who proposed a passive universal gripper based on a bistable-shell architecture. In their design, an axisymmetric shallow shell undergoes eversion upon contact with the target object, enabling compliant grasping arms to envelop the object without requiring an externally applied impact force. Unlike the design proposed by Zhang et al. [38], this mechanism demonstrated improved geometric adaptability by successfully grasping objects of varying sizes, shapes, and masses, including rings, paper clips, cylinders, miniature motors, and plastic spoons. The design integrates the switching mechanism (the bistable shell) and the compliant grasping arms into a single monolithic structure that can be 3D-printed, formed, or moulded, thereby eliminating the need for discrete joints while preserving the advantages of compliance-based systems such as passive actuation and zero power consumption in standby.

However, the inverting-shell mechanism still requires a localized mechanical force at the device to trigger bistable-state transitions between gripping and release. For implementation within a fully tetherless magnetic system, a non-contact switching approach is required, such as magnetically induced triggering of the established bistability. By embedding permanent magnets within a compliant bistable shell and driving state transitions using externally applied magnetic fields, the device can retain the advantages of passive, energy-efficient gripping while achieving remote, wireless control of its open and closed configurations [13].

Taken together, the four design constraints introduced above, together with

the compliant bistable design philosophy exemplified by prior work, directly motivate the conceptual architecture described in the following section.

2.2 Proposed Conceptual Design

The design constraints and passive bistable gripping principles outlined in Section 2.1 motivate a concrete device architecture that operates within the geometric and force limits of the colonic environment. Breaking the problem further down, the gripper platform comprises the following components:

1. **Bistable Shell:** A compliant bistable structure constructed as a segment of a hollow sphere or an equivalent curved geometry, responsible for the bistable open and closed states of the mechanism.
2. **Gripper Prongs:** Compliant grasping structures responsible for tissue engagement and retention, placed along the inner surface of the shell.
3. **Contact Pad:** An interface element used to trigger bistable state transitions, placed at the centre of the shell's inner surface. It contacts the target tissue feature before the prongs do.
4. **Magnetic Element:** A magnetic component responsible for device actuation, orientation, and rotational control, placed in a chosen arrangement along the rim of the shell.

The integration of these functional elements provides a conceptual basis for the design, as illustrated in Figure 2.2. To analyze and establish the design parameters and dimensional requirements of a practical tetherless bistable

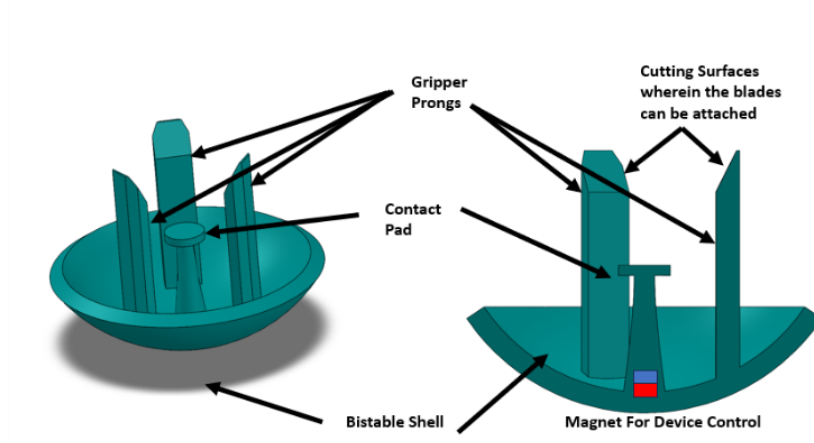


Figure 2.2: Proposed initial conceptual design of the tetherless bistable cutting device, showing the gripper prongs, contact pad, bistable shell, and magnetic element.

Table 2.1: Design constraints and dimensional requirements for the representative colonic target application, based on anatomical and robotic colonoscopy requirements reported by Suulker et al. [40].

Parameter	Requirement	Units
Overall device diameter	≤ 22	mm
Overall device length	≤ 22	mm
Required actuation force	≤ 1.5	N
Required gripping force	2–5	N

gripping device, a representative target application is first considered to help outline functional constraints. Considering the application to be polyps in the colon, the review by Suulker *et al.* [40] consolidates anatomical data reporting segment-specific dimensions ranging from 26 mm in the sigmoid colon to 45 mm in the ascending colon, thereby defining the lumen sizes within which the device must operate.

Based on published measurements of safe colon grasping forces and atraumatic laparoscopic grasper design, this thesis adopts a working gripping-force band of approximately 2–5 N for polyp engagement. Force levels in this range are consistent with instrumented studies in colon surgery, where safe graspers transmit up to about 5 N without macroscopic tissue damage, as noted by de Visser *et al.* [36], and are in line with the design of the force-adjustable parallel-occlusion grasper described by Khan *et al.* [37]. This force range is therefore expected to be sufficient to tent and stabilize polyps up to roughly 10 mm in size while maintaining atraumatic interaction with the surrounding colon wall. The MagnetoSuture system described by Mair *et al.* [13] provides an approximation for the maximum force applicable by an external magnetic system on an NdFeB needle to be approximately 1.34 N. Hence, the gripper should ideally actuate with a force of 1.5 N or less.

Together, these constraints define the key design targets summarised in Table 2.1: a device that fits within a 22 mm envelope, achieves a gripping force of 2–5 N, and requires no more than about 1.5 N of magnetic actuation to transition between its bistable states. These targets form the basis for the material selection, fabrication, and testing strategies described in Chapter 3,

in which specific compliant materials and manufacturing methods are chosen to realize the conceptual design in physical prototypes while meeting the constraints established here.

2.2.1 Manufacturing of Conceptual Design

Chapter 3 goes into the details of the choice of materials and the choice of stoichiometric ratio used for Silicone 00-30 Ecoflex for the fabrication of the conceptual design. It also goes into detail on the resin-printing-based manufacturing process. These sections discuss the additional manufacturing steps and takeaways after manufacturing the conceptual design.

2.2.1.1 Manufacturing using Silicone

The first step in the silicone manufacturing process is mold-making. The molds are created in SolidWorks, where the model is enclosed in a sufficiently large cuboid with a sufficient wall thickness. Split lines are created, and the cuboid forms a body cavity of the conceptual design. This results in a mold with 2 halves. To ensure alignment of the two halves, dowel-hole features are embedded in the mould halves, as seen in Figure 2.3. In Figure 2.3, when creating the split bodies, it is observed that the top half consists of 2 pieces, which are required to create a shell feature. These two pieces are superglued by carefully aligning the holes in the two molds to form the mold in Figure 2.4.

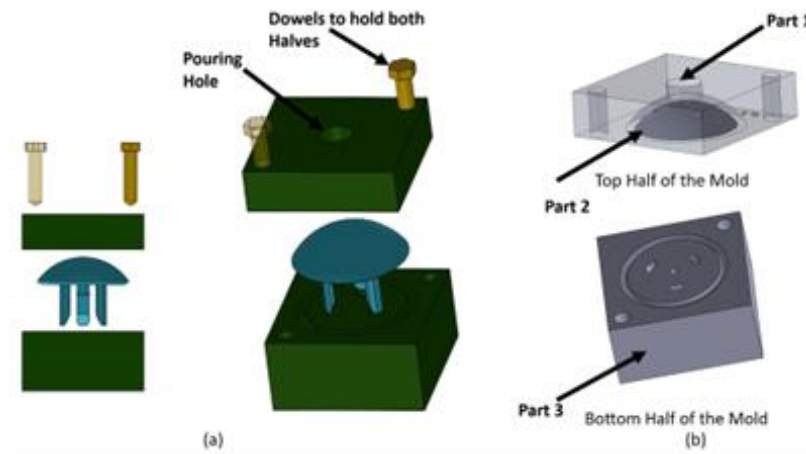


Figure 2.3: Mould Design in SolidWorks

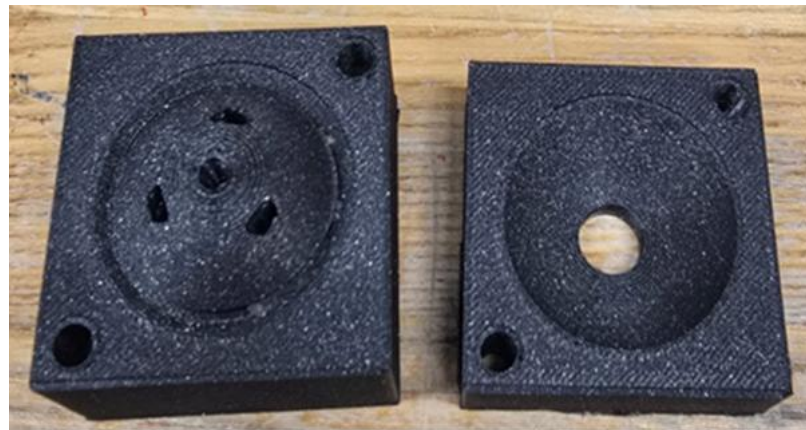


Figure 2.4: 3-D Printed Mould

After pouring the stoichiometric ratio of silicone determined in Chapter 3 and allowing it to cure for over 48 hours, the conceptual gripper was successfully extracted. One drawback noted was that the silicone grippers were too soft; therefore, separate moulds would be required for the shell and the gripper, which would then be assembled after curing. The curing process took more than 48 hours, with completion estimated by probing the excess silicone in the pouring hole. The extended curing duration was attributed

to the lack of oxygen within the enclosed mould, which hindered the curing process. In addition, the silicone had to be poured in multiple stages because it shrank as it cured. This two-part moulding and assembly process, combined with the long curing time and multi-stage pouring requirement, significantly increases design and manufacturing iteration time. Although silicone is a highly biocompatible material, the overall fabrication process was considered too time-consuming for this stage of development and is better suited for future work.

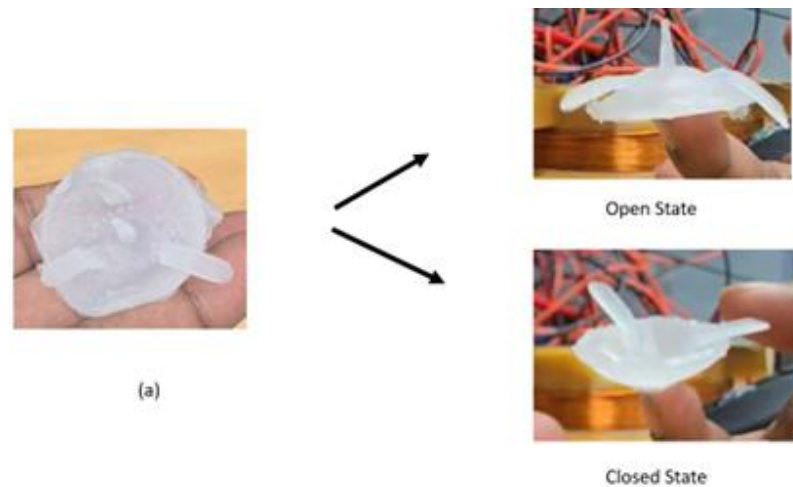


Figure 2.5: Fabricated Silicone Bistable Gripper

2.2.1.2 Manufacturing using Resin Printing

Figure 2.6 is the manufactured gripper through resin printing. Upon interacting with the gripper, it was observed that it was unable to maintain its bistable state. It was imperative to understand the parameters that affect the bistability of this design.



Figure 2.6: Resin-fabricated Gripper

2.2.2 Understanding the Bistability Nature of Conceptual Design through FEA

FEA study to understand the bistable nature is set up in CalculiX 2.21, using properties of the F80 resin from the manufacturer's data sheet. The resin material is modeled as a Neo-Hookean hyperelastic material, with $C_{10} = 1.54$ MPa and $D_1 = 0.053$ MPa⁻¹. In this Neo-Hookean formulation, C_{10} controls how stiff the resin is in shear ($\mu = 2C_{10} \approx 3.08$ MPa), and together with D_1 gives an effective $E_0 \approx 9$ MPa and $\nu \approx 0.46$, while D_1 sets how hard it is to compress the resin (through the bulk modulus, so smaller D_1 means the material behaves closer to incompressible). These constants are fixed a priori from the resin grade's data sheet before the start of this study and are held

identical across all iterations; they are never adjusted on a per-iteration basis to improve agreement with the experiment. The mesh uses C3D10 quadratic tetrahedral elements at a nominal edge length of 1.5 mm, with a reference case additionally verified at 0.8 mm to check for mesh sensitivity; a full mesh-refinement pass was not repeated across the entire iteration set. Node counts vary by iteration geometry, ranging from 13,708 to 44,830 (see Table 2.2). Because a simple static nonlinear solve breaks down at the snap-through limit point, the problem is instead run as a transient dynamic simulation: an implicit *DYNAMIC step with HHT- α time integration ($\alpha = -0.3$) is used so that the solution can pass smoothly through the snap-through event and capture the full force–displacement curve, including the unstable region.

The boundary condition for the FEA simulation is shown in Figure 2.7. The red mesh section is fixed, while the shell is subjected to a uniform load to induce deformation. This boundary condition is the one calibrated quantity in the model: it is calibrated to match the supported geometry (i.e., where the physical test rig constrains the shell), not to match the target force, and is applied unchanged across all iterations. This distinction matters because a per-iteration-fitted model could not, by construction, produce the rows discussed below where no snap-through is predicted at all; a genuinely re-tuned model would trivially reproduce a snap in every case.

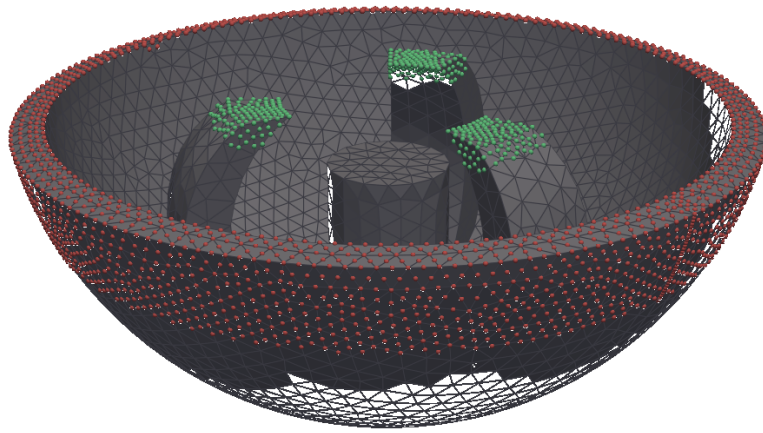


Figure 2.7: FEA Boundary Conditions

Running the FEA simulation yields the results as shown in Figures 2.8 and 2.9.

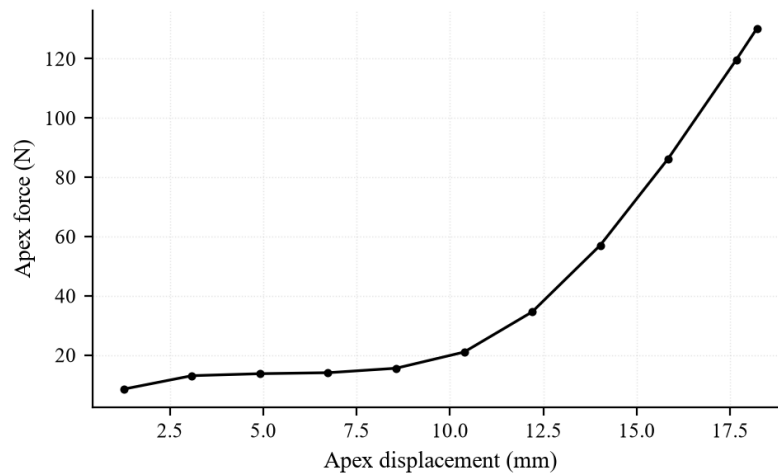


Figure 2.8: Conceptual Design Force vs Displacement

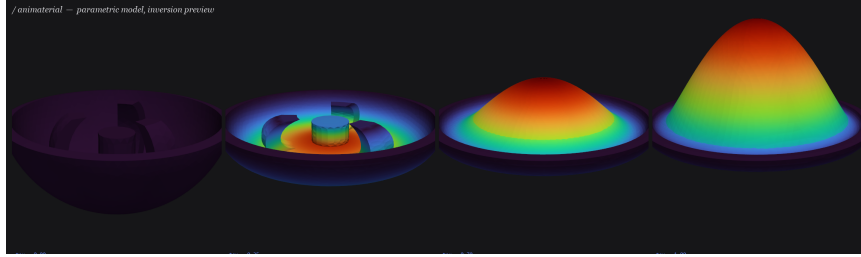


Figure 2.9: Mesh Deformation Upon Force Application

The Graph in Figure 2.8 shows that there is no bistable point for the geometry used to design the bistable shell in the conceptual design.

The first step is to determine whether the shell profile significantly affects the shell's stiffness. The shell profiles are modified in accordance with Figure 2.10.

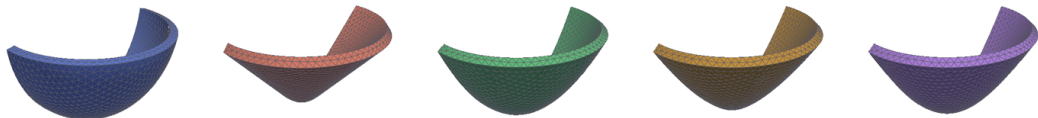
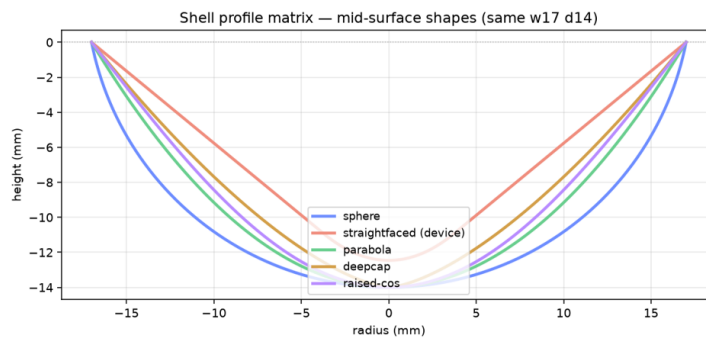


Figure 2.10: Shell Profiles

Running a parametric sweep yields the results shown in Figure 2.11. The key takeaways from these curves are that we would like a lower snap-through force. From the graph, a sphere would be ideal, but it would have limited

practical applications for our conceptual design; hence, the next most ideal shape is utilised for further designs. Here, we also ensure that the shell's width and depth remain constant across profiles.

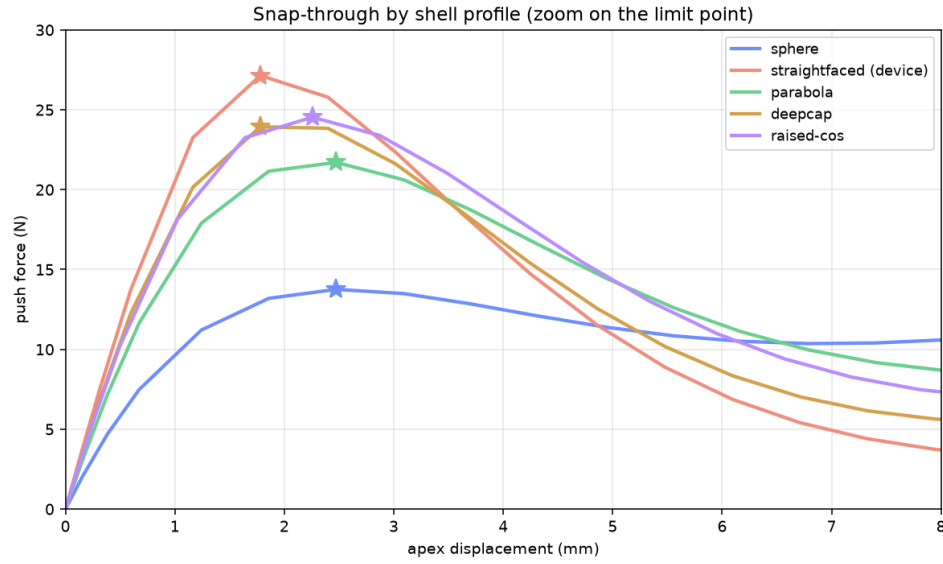


Figure 2.11: Snap through for Shell Profile

To further deconstruct the design parameters, the bistable shell of the gripper is defined using the parameters shown in Figure 2.12.

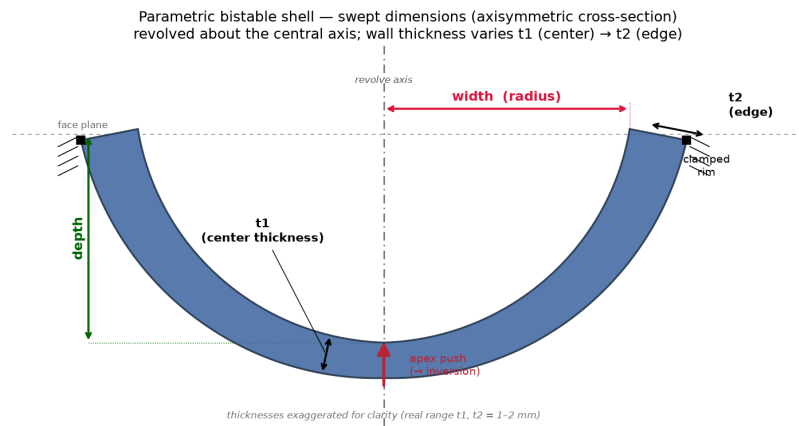


Figure 2.12: Parameters utilised to define the shell bistability

The first parameter, "width," is defined as the distance from the shell's central axis to the inner edge. It is hypothesised that as the parameter increases geometrically, the shell diameter increases, reducing the force required to invert between the two states. An assumption made here is that the shell thickness is constant. The second parameter, "depth," is defined as the distance from the horizontal axis on the shell's face to its lowest point. This controls the overall depth of the shell; it is hypothesised that as this parameter increases, the force required to invert the shell between the two states increases. Similar to the first parameter, this assumes a constant shell thickness. Now the parameters "t1" and "t2" are represented in Figure 2.12. The ratio between these parameters gives rise to the 3 cases as shown in Figure 2.13.

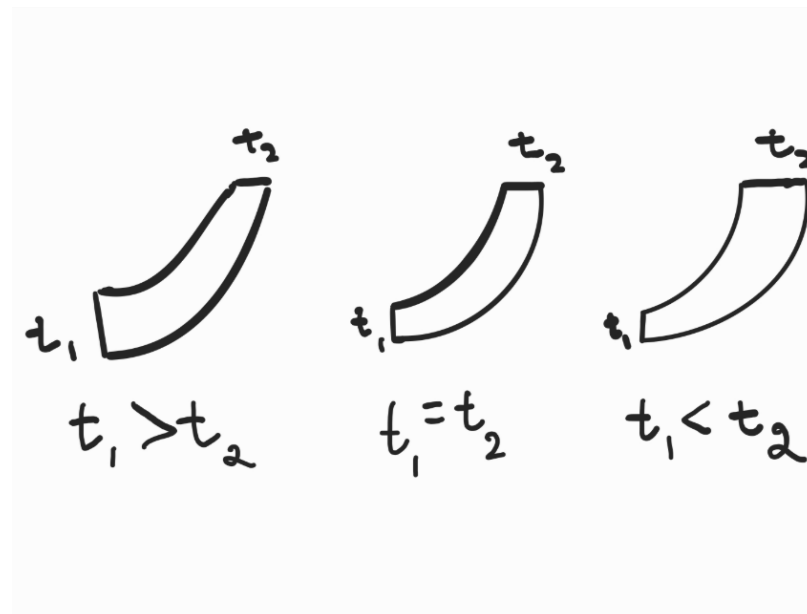


Figure 2.13: Parameters ratio edge cases

Running a parametric sweep on these dimensions, the obtained results are present in Figures 2.14, 2.15, 2.16, and, 2.17.

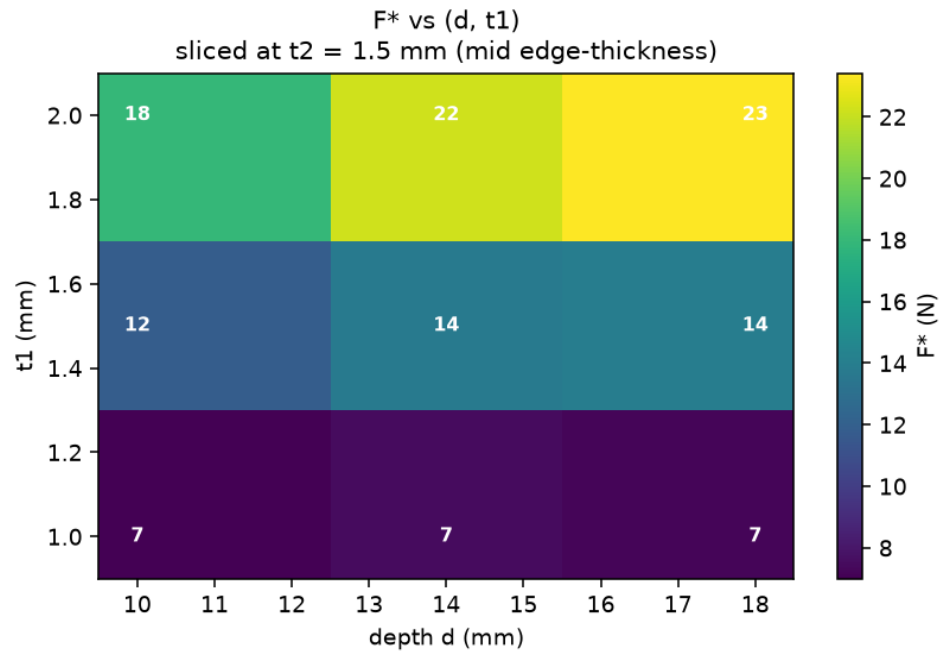


Figure 2.14: Parameter 'Depth vs t_1 '

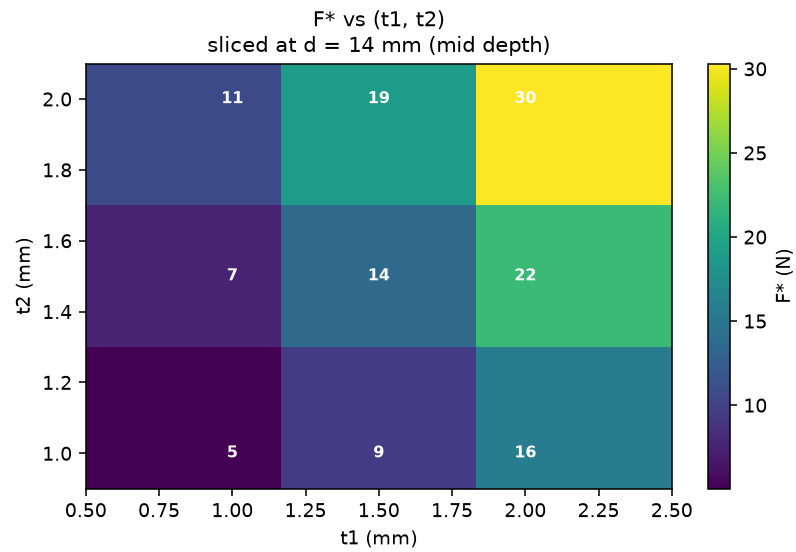


Figure 2.15: Parameter ' t_2 vs t_1 '

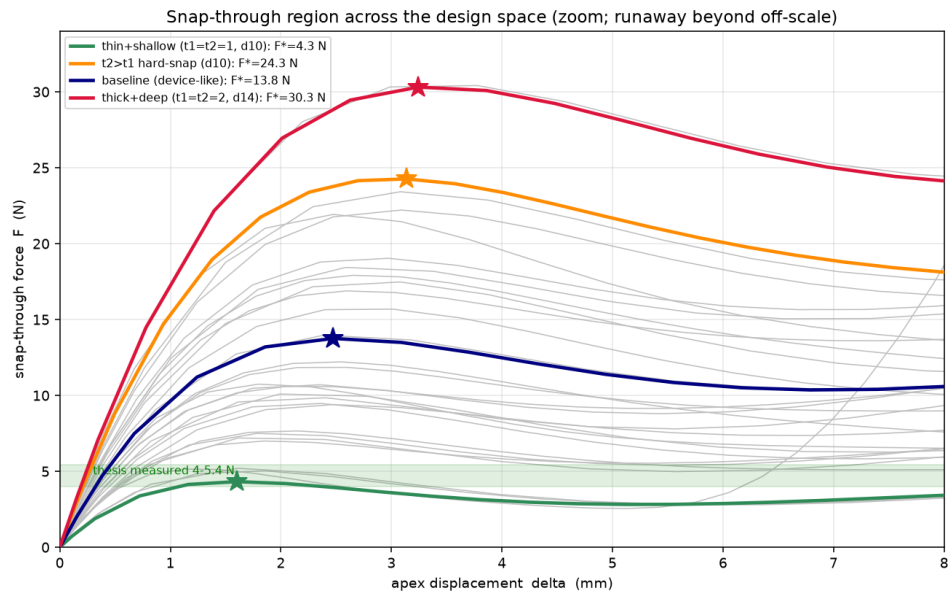


Figure 2.16: Snap-through across the design Spaces

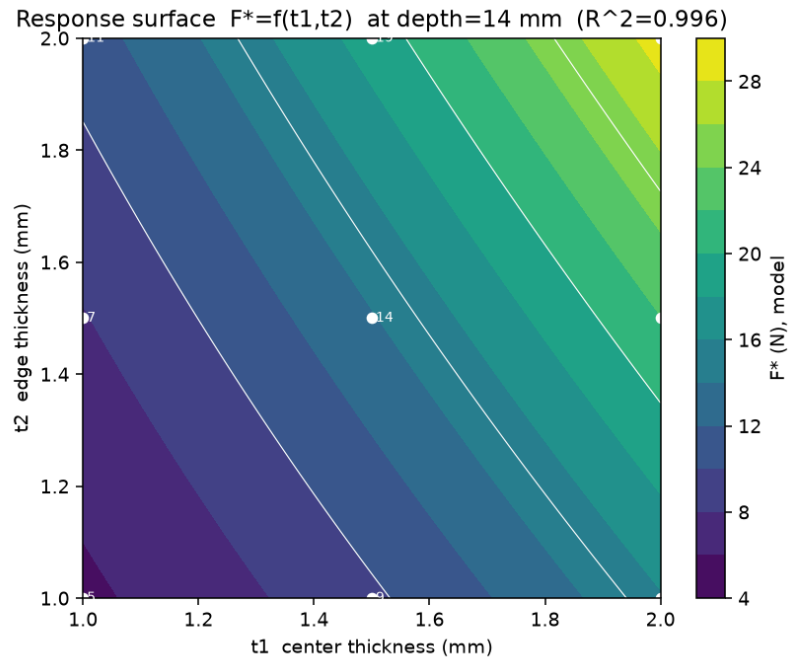


Figure 2.17: Parameter 't1 vs t2' design Space Family of Curves

Table 2.2 at the end of this section summarizes the fixed FEA settings used throughout this study.

Quantity	Value	Reasoning
Material model	Neo-Hookean hyper-elastic	Chosen for large-strain snap-through behavior
C_{10} (Shear Stiffness Parameter)	1.54 MPa	Fixed from the F80 resin grade data sheet before the study; never adjusted
D_1 (Compressibility Parameter)	0.053 MPa^{-1}	Fixed from the F80 resin grade data sheet before the study; never adjusted
Initial modulus E_0	$\approx 9 \text{ MPa}$	Calculated through C_{10} and D_1
Poisson's ratio	≈ 0.46	Calculated through C_{10} and D_1
Element type	C3D10 (quadratic tetrahedra)	–
Nominal edge length	1.5 mm	Reference case additionally verified at 0.8 mm; no full mesh-refinement pass across the set
Node count range	13,708 – 44,830	Varies by iteration geometry
Time integration	Implicit *DYNAMIC, HHT- $\alpha = -0.3$	Static Newton–Raphson solve diverges at the limit point
Boundary condition	Fixed support region, uniform applied load	Calibrated to supported geometry, not to force; applied unchanged across all iterations
Solver	CalculiX 2.21	Two independent builds agree to $\sim 1\%$

Table 2.2: Summary of FEA method and settings used throughout this study.

Chapter 3

Material Selection, Manufacturing & Testing

The conceptual design in Chapter 2 utilizes a bistable shell as an integral component. Bistability can be achieved mainly through two methods. The first method is straightforward: it uses a compliant material; the second method modifies the geometry using a specific topology to achieve the required compliance. In the case of the gripper, a hybrid approach was adopted, in which an already compliant material was used, and the geometry was modified to achieve the required target. This chapter of the thesis discusses the process of choosing a compliant material for our application. It also covers the testing methodology used to characterize the manufactured devices and meet the requirements stated in Table 2.1.

The available options for fabricating flexible structures within the available laboratory facilities included two-part silicone rubber casting and resin-based additive manufacturing using TPU-like flexible materials. Although flexible filaments are available for conventional fused deposition modeling (FDM), this option was not considered due to the complex geometry of the proposed

design and the extensive post-processing required.

3.1 Experimenting with Silicone Resin

The final mechanical properties of silicone structures are strongly influenced by both the stoichiometric mixing ratio and the preparation procedure. To evaluate the influence of composition on curing behavior and material stiffness, a standardized mould was fabricated for preliminary characterization experiments.

Smooth-On EcoFlex 00-30 silicone rubber was selected as the base material. A range of Part A to Part B mixing ratios, varying from 10:90 to 90:10, was investigated, as shown in Figure 3.1.



Figure 3.1: Mold used to evaluate different stoichiometric ratios of two-part silicone.

Experimental observations indicated that samples prepared with lower Part A ratios (10:90, 20:80, and 30:70) did not fully cure within the mold geometry. Ratios exceeding 40:60 exhibited stable curing behavior and maintained structural integrity after demolding. In addition, the stiffness of the cured silicone increased approximately linearly with increasing Part A concentration.

However, mixtures with higher Part A ratios expanded beyond the mould boundaries during curing, indicating a need to accommodate overflow in the

mould design. To address this issue, spill-over ridges were incorporated into the mould geometry, allowing controlled silicone expansion during the curing process.

The curing characteristics also differed significantly between open-air exposure and enclosed mould conditions. During curing, the manufacturer-recommended 24-hour curing period proved insufficient; complete curing required more than 48 hours. Air entrapment during mixing additionally introduced internal voids and inconsistencies in the final structure. These observations suggest that vacuum degassing could further improve fabrication quality and repeatability by minimizing trapped air bubbles.

Based on the observed curing behavior, structural consistency, and mechanical stiffness, a stoichiometric ratio of 70:30 was selected as the optimal composition for fabrication of the proposed device.

3.2 Manufacturing Using F80 in a Resin Printer

The resin-based manufacturing process began by slicing the STL model using dedicated slicing software and transferring the generated file to the printer via a USB interface.

Following printing, several post-processing operations were required to achieve the desired print quality and dimensional accuracy. The first stage involved cleaning the printed components in an isopropyl alcohol (IPA) bath with ultrasonic agitation to remove excess uncured resin from the surface. This was followed by ultraviolet (UV) curing, during which the components were exposed to controlled UV illumination on a rotating platform to ensure

uniform hardening of the polymer.



Figure 3.2: Resin Manufacturing Process

Although these post-processing procedures are standard in resin-based additive manufacturing, several process-specific observations were identified during fabrication.

It was observed that immersing the printed component in the IPA bath before support removal significantly improved part reliability and reduced the likelihood of structural damage. This improvement resulted from the difficulty in distinguishing support boundaries before complete resin cleaning. The part was also more malleable before cleaning and the resin hardening process, and was therefore prone to tearing during handling. Experimental optimization indicated that an IPA cleaning duration of approximately 20 minutes, with orientation changes every 10 minutes, was effective. It was also observed that placing the prints in a particular orientation improved cleaning performance. Similarly, UV curing for approximately 21 minutes,

with periodic orientation changes every 7 minutes, resulted in more uniform curing.

Additional manufacturing inconsistencies were traced to printer maintenance conditions. Degradation or improper replacement of the release film significantly affected print quality and surface finish. Furthermore, inadequate cleaning of the washing bath allowed residual uncured resin to adhere to printed parts during post-processing, resulting in unintended hardened artifacts on the final components.

These manufacturing defects are illustrated in Figures 3.3 and 3.4.



Figure 3.3: Post-curing defects observed after UV curing.



Figure 3.4: Printing defects observed during resin-based manufacturing.

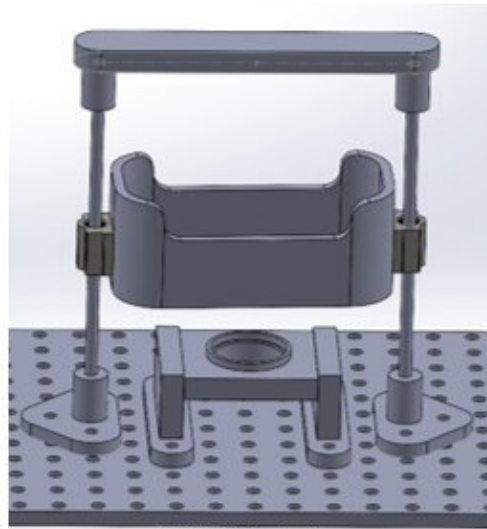
3.3 Testing

Two setups were manufactured to characterize actuation and gripping forces, as shown in Table 2.1. This section discusses the setup's manufacturing and the gripper's characterization.

3.3.1 Actuation Force Characterization

To characterize the actuation force across multiple iterations, a reliable method for testing the bistable shell is needed. To achieve this, a simple test setup inspired by a UTM was designed and manufactured, as shown in Figure 3.5. The test rig is mounted on an optical table to ensure rigidity and levelness.

The testing procedure is to place the shell in its first stable state, then place the weight in the weight holder to apply a load. The weight is then removed to see if the shell flips into its second stable state. Figure 3.7 shows



SolidWorks Design



Manufactured Force
Testing Mechanism

Figure 3.5: Test rig for bistable shell force characterization

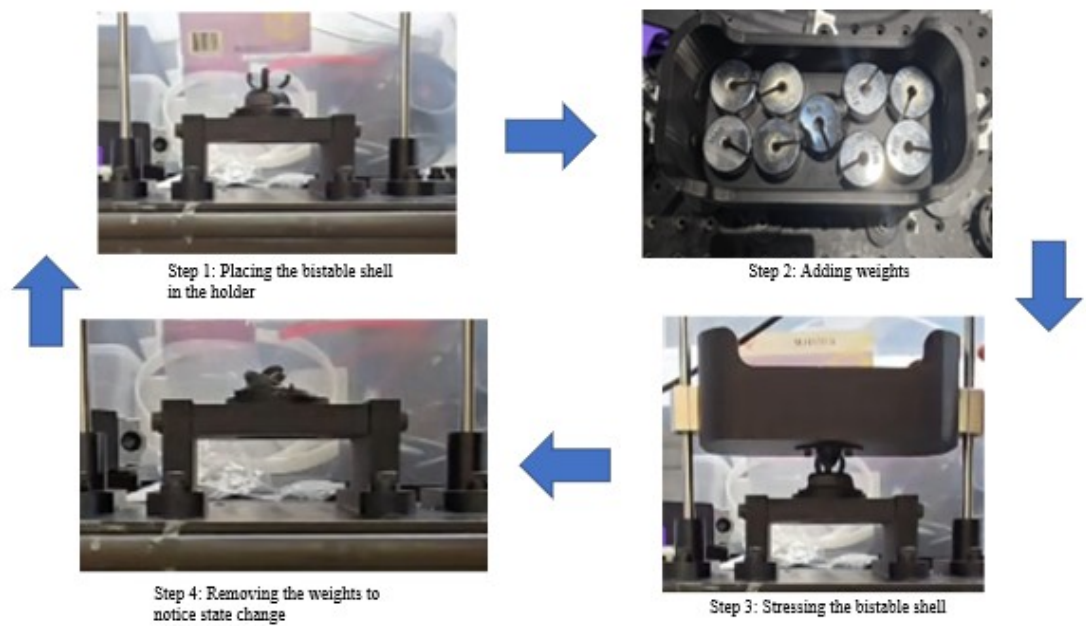


Figure 3.6: Testing Procedure

the bistable shell holder, with two significant features. The first feature would be the addition of a rim, which would allow for the shell to be constrained in the axial direction. The second feature would be a cut-out hole in the shell holder. This is designed to remove the ground reaction forces the shell would experience when placed on a flat surface. There is a small lip that would hold the shell's rim. This is a limitation, but it has been accounted for in the FEA simulation. The weight holders have two iterations: the first was found to

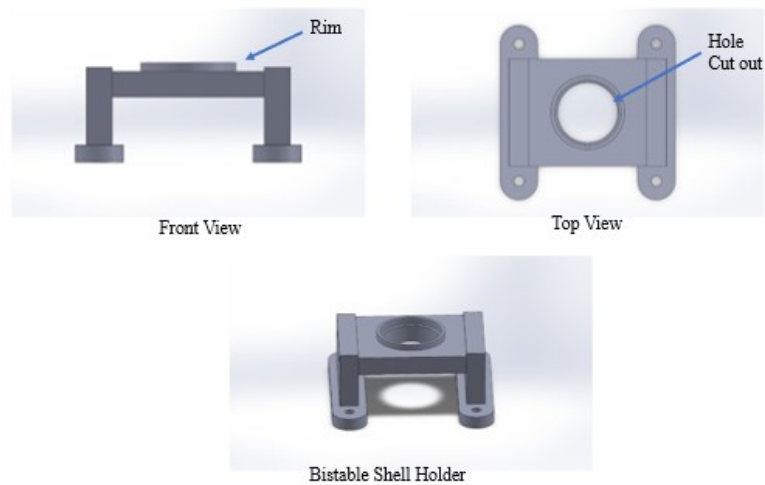
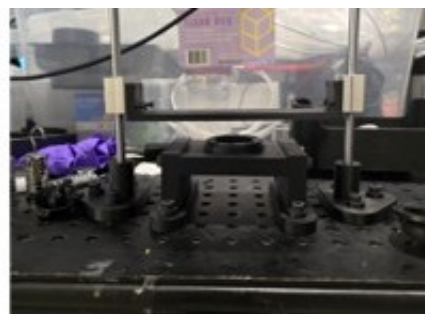


Figure 3.7: Bistable Shell holder

be too heavy and would trigger additional shell iterations even without any other slotted weights. Hence, it was limiting the characterization of the shell, so a lighter-weight holder was manufactured, as shown in the Figure 3.8. The second iteration of the weight holder was 165 grams lighter. One of the drawbacks of the current machine is that the weight holder occasionally gets stuck on the rails, which could be improved by replacing them with linear rails. Instead of 3D printing the pieces, manufacturing them with metal would



Weight Holder Version 1



Weight Holder Version 2

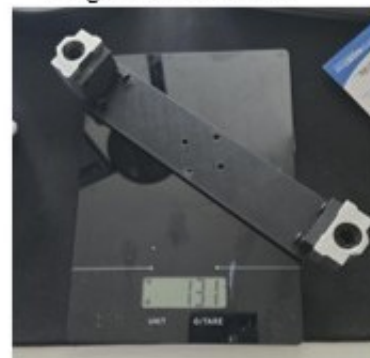


Figure 3.8: Weight reduction between the iterations

have better tolerances.

3.3.2 Test-rig for Gripper Strength Characterization

Figure 3.9 shows the manufactured test rig for finding the gripping strength of the bistable gripper devices. The plate is suspended from the edge of the table so the weights can hang. The cantilever effect of the test rig is negligible due to the plate's rigidity, size, and the range of weights used to test the gripper's strength. The plate still uses the same design philosophy as the plate shown in Figure 3.7.

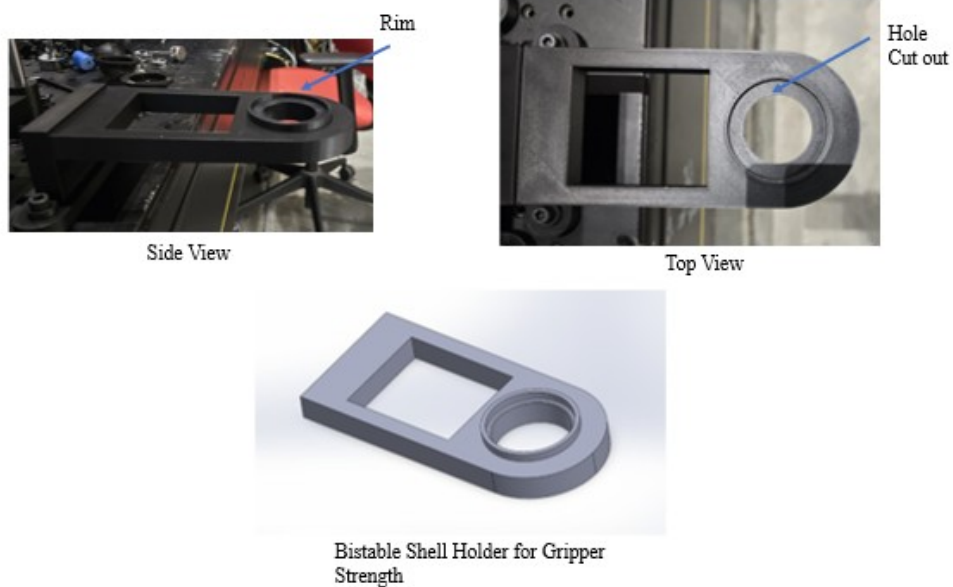


Figure 3.9: Test-rig for Gripper Strength Characterization



Bistable Gripper holding a polyp phantom



Bistable Gripper loaded with slotted weights

Figure 3.10: Testing Procedure for Gripper Force

Chapter 4

Design Iteration

4.1 Summary

This chapter documents the iterative mechanical redesign of the bistable gripper shell, from the initial conceptual geometry (Iteration 1) through seven subsequent design revisions (Iterations 2–7). Each iteration was evaluated using the two complementary methods established in Chapter 3: (1) experimental snap-through force characterization on the custom UTM-inspired test rig (Section 3.3.1, Figures 3.5–3.8), in which the shell is loaded in triplicate to its snap-through point and the average force recorded; and (2) finite element analysis (FEA) in CalculiX using a Neo-Hookean hyperelastic model of the F80 resin (Section 2.2.2), which predicts the snap-through force from the shell’s geometric parameters.

Table 4.1 summarizes the experimental versus FEA snap-through forces for all design iterations, and Figure 4.1 overlays the corresponding force vs displacement curves. Throughout the design progression, the experimentally measured actuation force decreased from 14.23 N (Iteration 2 baseline) to

1.93 N (Iteration 7), a cumulative reduction of approximately 86%. The primary design changes driving this reduction were: introduction of a flexure-like feature (Iteration 3), fillet-based mitigation of stress concentrations (Iteration 4), magnet-lip minimization and arc-section redesign (Iteration 5a/Iteration 5b), and depth tuning followed by material removal from the shell quadrants (Iterations 6 and 7).

Experiment-to-FEA agreement was generally within 3–15% absolute error, with the notable exceptions of Iterations 6 and 7 (37.2% and 15.6% error, respectively), where the physical devices were composite assemblies consisting of an F80 resin shell bonded to rigid PLA gripper prongs and contact pads. In contrast, the FEA model treated only the shell as a homogeneous hyperelastic body. The implications of this modeling choice are discussed in Sections 4.6 and 4.7.1. Gripper-prong performance across four candidate designs and multiple phantom geometries is summarized in Tables 4.2 and 4.3 and discussed further in Section 4.6.1.

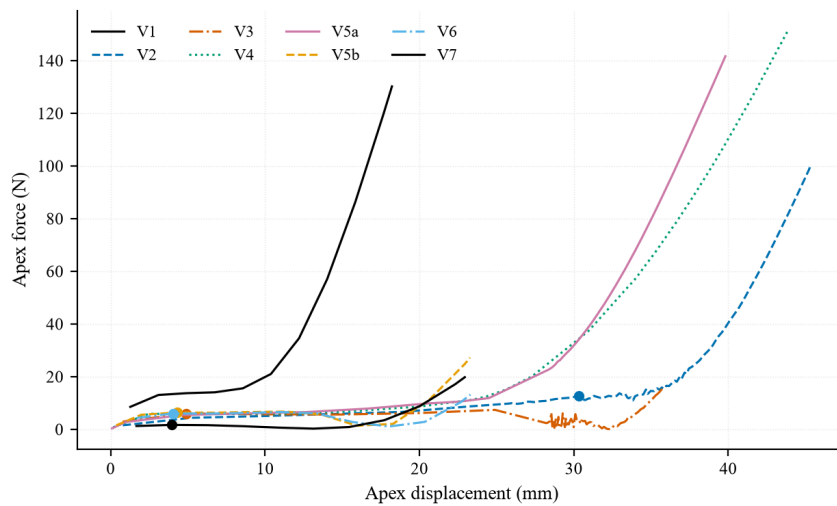


Figure 4.1: Force–displacement snap-through curves for all iterations.

Iteration	Trial 1	Trial 2	Trial 3	Average (N)	FEA (N)	Error (%)
Iteration 1	–	–	–	–	No snap-through	–
Iteration 2	14.0	14.5	14.2	14.23	12.66	12.4
Iteration 3	6.5	6.8	7.0	6.77	6.00	12.8
Iteration 4	7.5	9.0	8.3	8.27	8.00	3.4
Iteration 5a	12.2	13.0	12.5	12.57	11.70	7.4
Iteration 5b	6.7	7.3	7.8	7.27	6.36	14.3
Iteration 6	4.0	3.5	3.8	3.77	6.00	37.2
Iteration 7	1.9	2.0	1.9	1.93	1.67	15.6

Table 4.1: Experimental snap-through forces versus FEA predictions across design iterations. Error is computed as the absolute percentage difference, $\text{Error} = \left| \frac{\text{Average} - \text{FEA}}{\text{FEA}} \right| \times 100\%$.

4.2 Iteration 2

There are two main changes between the Conceptual Design (Iteration 1) and Iteration 2. The first is the addition of a section to house the permanent magnets that control the device, and the second is a change in shell dimensions to achieve bistability. While this design achieved bistability, the selected dimensions led to numerous stress concentration points in the shell. Under repeated manual actuation, the shell was prone to tearing along these high-stress regions.

Another observation was that, in this iteration, the gripper arms would catch on the inner lip of the rim, requiring more force to switch states and occasionally becoming completely stuck. Similar behavior was observed in the FEA snap-through curve, shown in Figure 4.2, where the force plateau is broad, and the snap-through point occurs at a relatively high load. Using the actuation-force test rig and testing procedure described in Section 3

(Figures 3.5–3.7), triplicate trials produced an average snap-through force of 14.23 N (Table 4.1), against an FEA-predicted value of 12.66 N (12.4% absolute deviation).

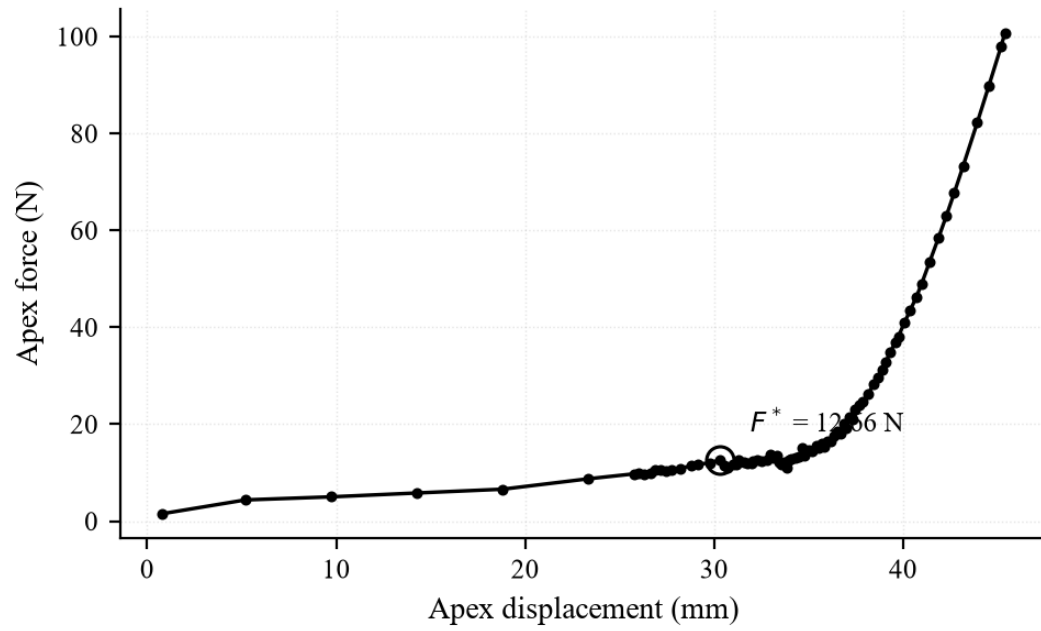


Figure 4.2: Force vs displacement for Iteration 2.

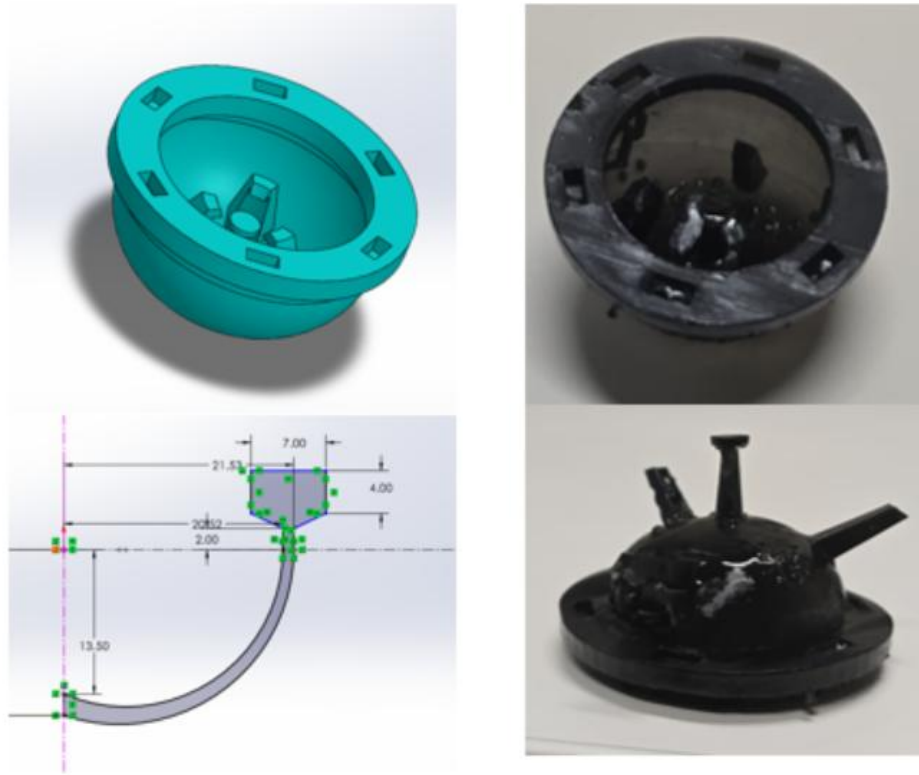


Figure 4.3: Iteration 2: Top Left (gripper in SolidWorks), Top Right (closed state), Bottom Left (shell cross-section), Bottom Right (open state).

4.3 Iteration 3

Between Iterations 2 and 3, the rim geometry was modified to reduce interference between the rim and the gripper fingers. The orientation of the magnet holder was also changed, requiring further adjustment to the rim dimensions. The most significant change in this iteration was the addition of a flexure-like section in the shell, intended to reduce the actuation force required to trigger snap-through.

This effect is corroborated by the FEA results shown in Figure 4.4, where

the flexure feature yields a predicted snap-through force of 6.00 N, representing a reduction of approximately 52% relative to Iteration 2's FEA baseline of 12.66 N. Applying the same actuation-force test-rig methodology from Section 3, triplicate experimental trials gave an average snap-through force of 6.77 N (Table 4.1), a 12.8% absolute deviation from the FEA prediction of 6.00 N. These results confirm the flexure's force-reducing effect experimentally and computationally, but pose manufacturing drawbacks.

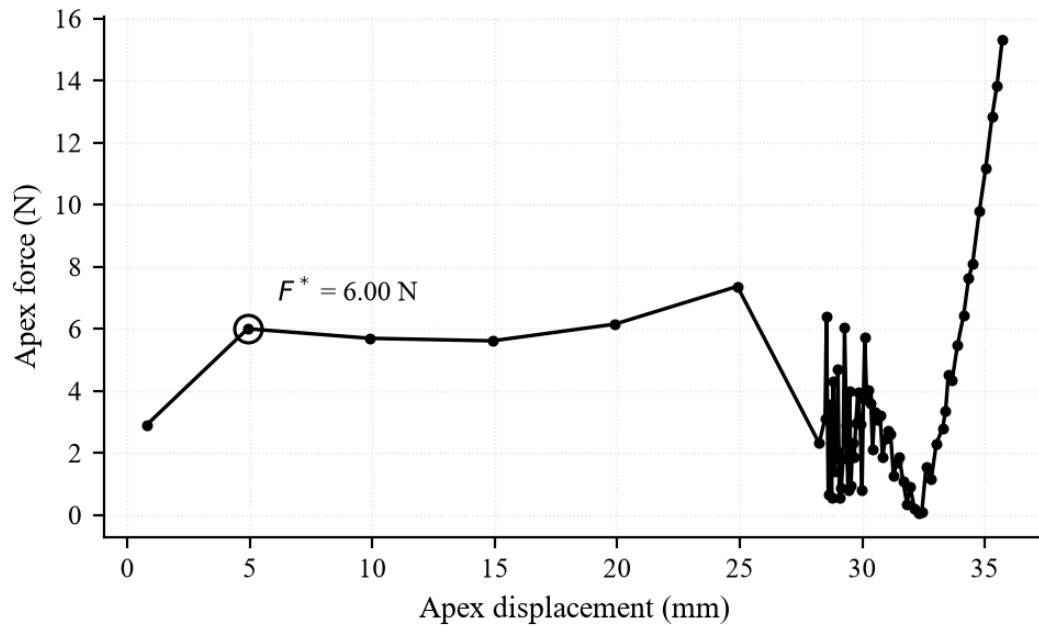


Figure 4.4: Force vs displacement for Iteration 3.

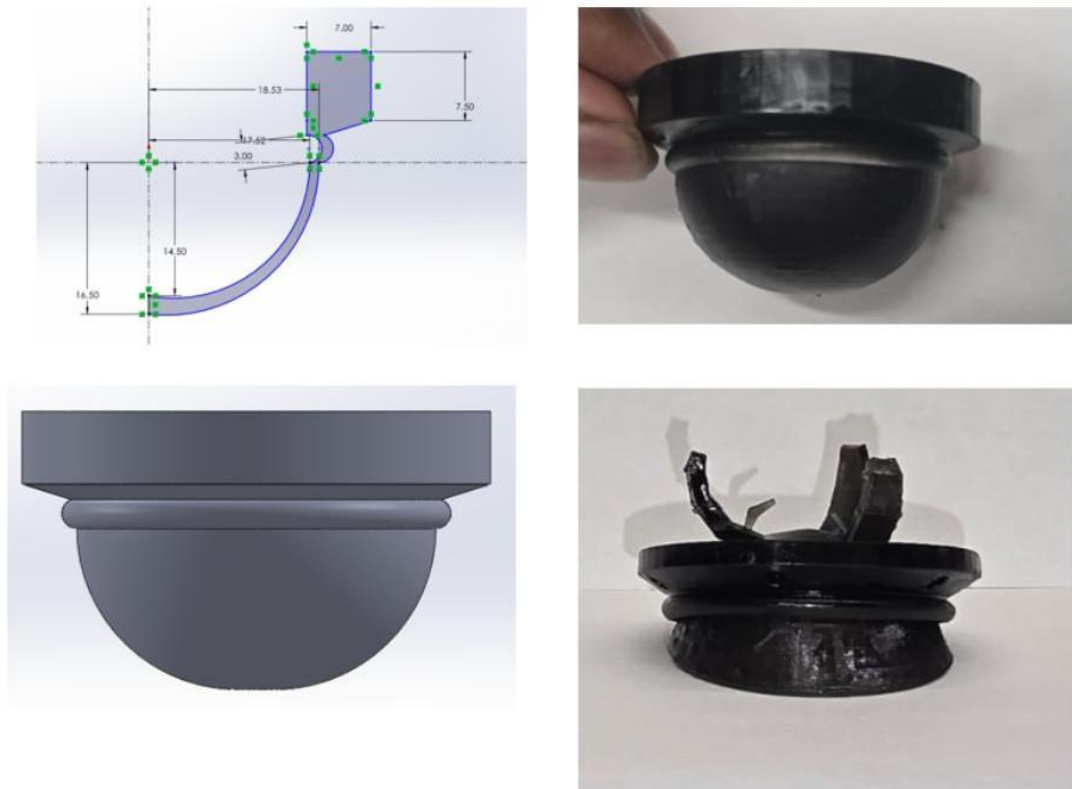


Figure 4.5: Iteration 3: Top Left (shell cross-section), Top Right (closed state), Bottom Left (gripper in SolidWorks), Bottom Right (open state).

While these results were ideal in simulation and initial testing, the failure mode observed under repeated actuation is shown in Figure 4.6. The flexure, while allowing snap-through under approximately 7 N of force on the test rig, acted as a stress-concentration line, inducing a shearing force that separated the rim from the rest of the bistable shell. The snap-through point force in this configuration was approximately 6.77 N (Table 4.1), but the associated structural reliability was insufficient for repeated actuation.



Figure 4.6: Shearing of the rim and bistable shell in Iteration 3 due to stress-concentration points along the flexure.

4.3.1 Magnet Orientation and Preliminary Proof of Control

4.3.1.1 MagnetoSuture System & Control

The electromagnetic actuation system used in this work follows the configuration proposed by Mair et al. [13]. Four coils are arranged orthogonally along the X and Y axes of a planar workspace to generate steerable magnetic field gradients that produce translational forces and alignment torques on embedded permanent magnets in the device. A fifth coil, positioned perpendicular to this planar arrangement, provides a controllable downward magnetic force for out-of-plane actuation.

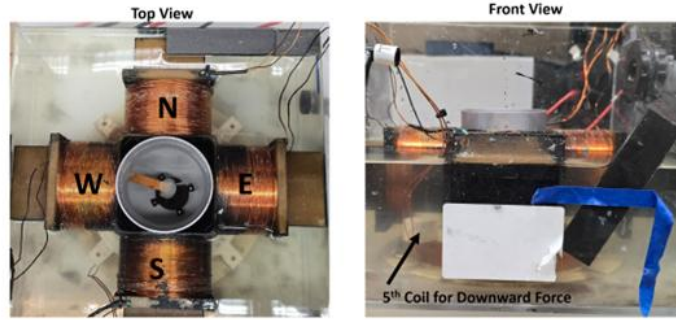


Figure 4.7: MagnetoSuture-inspired electromagnetic coil system.

Control signals are organized into three hierarchical layers (Figure 4.8). MATLAB serves as the supervisory layer, computing coil current commands from joystick input or predefined actuation sequences and transmitting them to an Arduino UNO via USB-serial. The Arduino converts these commands into PWM outputs for the motor drivers, which are powered by a dedicated DC supply and convert duty cycles into proportional coil currents. A safety constraint, enforced in both the firmware and MATLAB, keeps the sum of the squared normalized coil currents below unity to prevent excessive current loading.

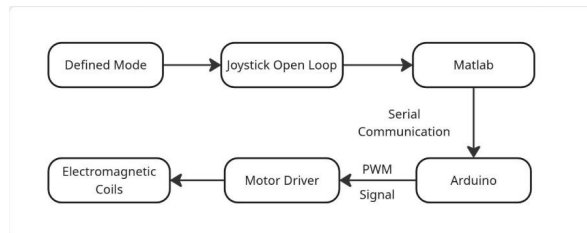


Figure 4.8: Open-loop control flow diagram for the coil actuation system.

Each coil is wound with AWG 16 polyimide-coated copper wire on a 3D-printed ULTEM 1010 frame (DC resistance $\approx 2.7 \Omega$, 20 A maximum current). A single energized coil with a soft-iron core produces 46.7 mT at a working

distance of 50 mm; activating two opposing coils raises this to approximately 93.4 mT, yielding an average field gradient of 2,778 mT/m within 20 mm of the coil face. Guided by the tissue-penetration measurements reported in the MagnetoSuture system [13], and using conservative magnetic-force estimates. This coil platform was designed to deliver a maximum force on the order of 1.34 N on an appropriately sized NdFeB element. This value sets the approximate upper bound of the snap-through force that can be magnetically triggered in the present device configuration.

4.3.1.2 Magnet Orientation and Selection

The device embeds a single set of perpendicularly oriented permanent magnets, chosen to support both bistable gripping actuation and straight-line translational motion simultaneously. In this configuration, the horizontal field component from the planar coil array couples to the magnet, enabling controlled 2D planar movement, while the vertical field component from the downward coil induces out-of-plane bistable-state transitions required for gripping.

Preliminary bench-top testing in air demonstrated that this perpendicular magnet arrangement can produce both translational motion and bistable flipping using the same magnet set, and that the available forces are sufficient to move the robot within a 2D plane under magnetic actuation. These results represent a proof of concept for the actuation strategy rather than a fully characterized locomotion system. Further testing and design optimization will be required to characterize performance in a liquid environment, where

fluid drag, buoyancy, and viscous damping are expected to alter the magnetic force and torque requirements needed for reliable actuation and gripping.

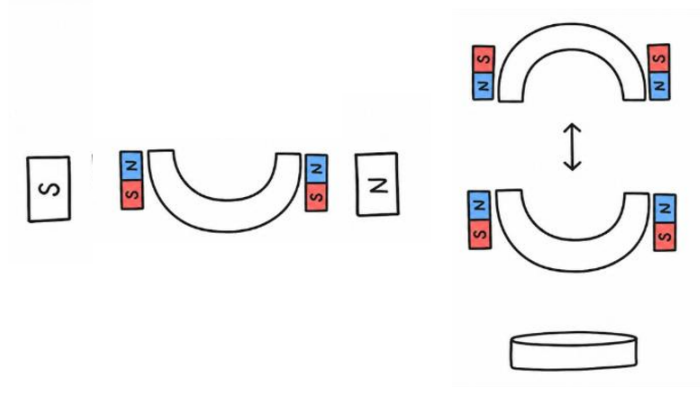


Figure 4.9: Magnet orientation for gripper control, showing perpendicular magnet configuration.

4.4 Iteration 4

In Iteration 4, the stress-concentration problem observed in Iteration 3 was addressed using standard design practices, such as adding fillets to smooth geometry transitions. Another design choice was to reduce the lip that houses the magnets, allowing them to sit more externally and thereby reducing mechanical interference with the bistable shell's deformation.

These changes yielded satisfactory results in structural robustness: the bistable shell maintained its state and did not fracture under repeated actuation. However, they introduced a new issue: the actuation force increased relative to Iteration 3. This is consistent with the parametric analysis presented in Section 4.3, where an external fillet and the resultant $t_2 > t_1$ condition increase the effective stiffness of the shell. As a result, the overall force required to trigger snap-through rose.

This behavior is corroborated through testing and FEA. Figure 4.11 shows that the force–displacement curve exhibits a shallow plateau rather than a sharp transition point, and the snap-through occurs at a higher force than in Iteration 3. Using the same test-rig protocol from Section 3, the average experimental actuation force across triplicate trials was 8.27 N, closely matching the FEA-predicted value of 8.00 N (3.4% absolute error, the tightest agreement of any iteration in Table 4.1). While Iteration 4 required a greater actuation force than Iteration 3, it successfully addressed the major structural failure point of Iteration 3 (rim shearing), making the force penalty an acceptable trade-off at this stage of the design.

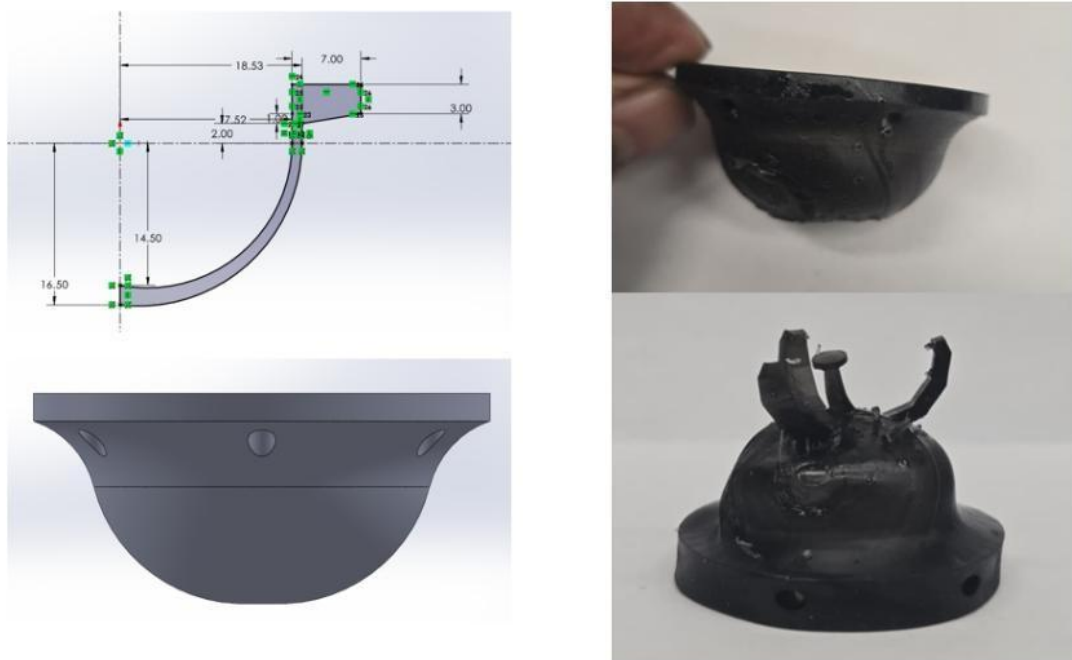


Figure 4.10: Iteration 4: Top Left (shell cross-section), Top Right (closed state), Bottom Left (gripper in SolidWorks), Bottom Right (open state).

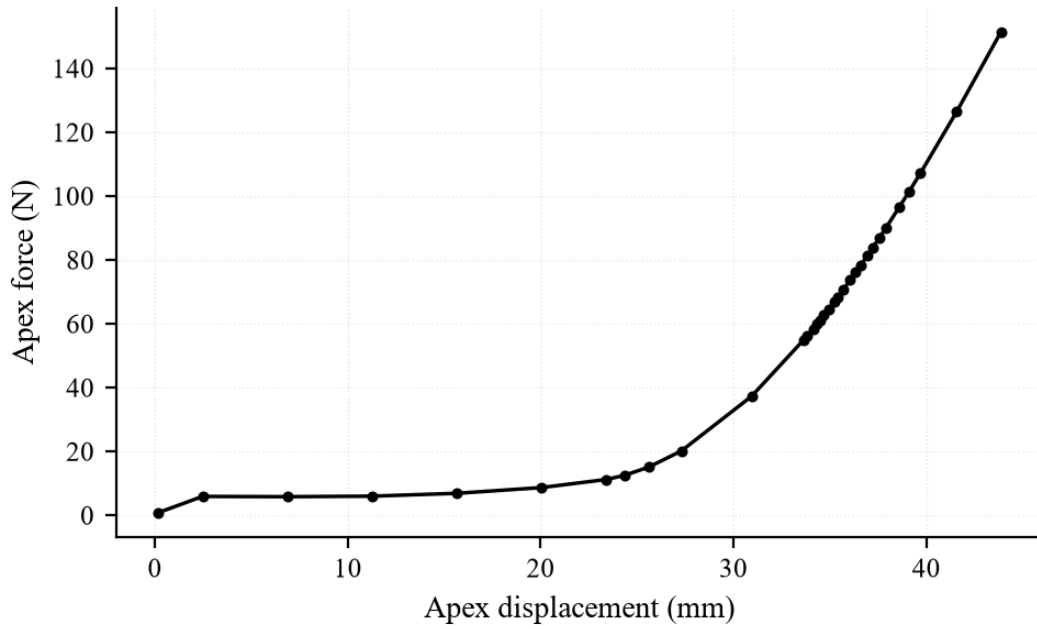


Figure 4.11: Force vs displacement for Iteration 4.

4.5 Iteration 5

4.5.1 Iteration 5a

Building on the parametric insights from Section 4.3, the source of the elevated actuation force was identified as the imbalance between thickness parameters t_1 and t_2 , specifically the condition $t_2 > t_1$ associated with the full magnet-holding lip. Iteration 5a focuses on minimizing this effect without altering the shell's overall curvature profile. The magnet lip was reduced to discrete cavities for the magnets to be trapped between, as shown in Figure 4.12, thereby reducing the effective width of t_2 .

Using the actuation-force test rig from Section 3, triplicate trials gave measured snap-through forces of 12.2 N, 13.0 N, and 12.5 N, for an average

of 12.57 N (Table 4.1), against an FEA prediction of 11.70 N (7.4% absolute error). While these results agree reasonably well with FEA, the actuation force remained much greater than Iteration 3's value of 6.77 N, indicating that reducing the magnet lip to cavities alone increased local rigidity and did not substantially lower the actuation force. Critically, however, the device retained the structural integrity gained from Iteration 4's fillet improvements, making Iteration 5a the first configuration in this study to be simultaneously bistable, structurally durable under repeated actuation, and compatible with magnet retention—albeit with a limited reduction in force relative to Iteration 4.

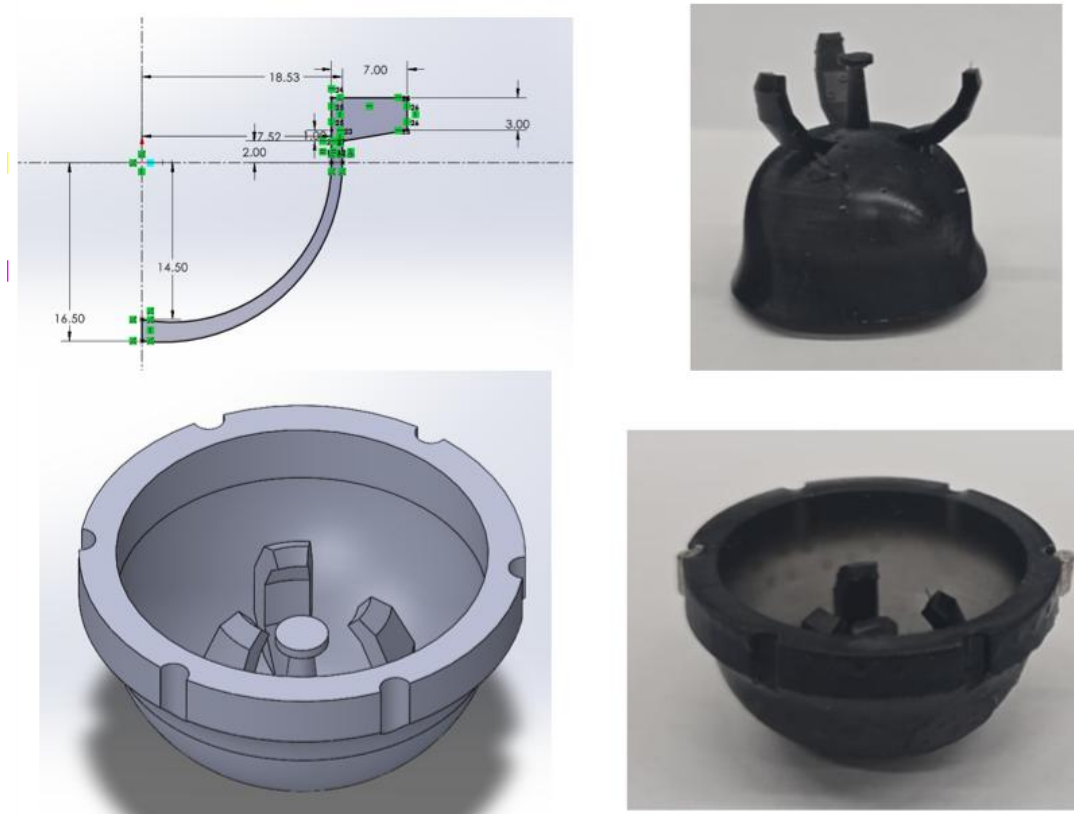


Figure 4.12: Iteration 5a: Top Left (shell cross-section), Top Right (closed state), Bottom Left (gripper in SolidWorks), Bottom Right (open state).

4.5.2 Iteration 5b

The second approach was to remove a larger portion of the lip while still providing enough arc sections to hold the magnets, as shown in Figure 4.13. Four evenly spaced arc sections were sized to preserve the shell's snap-through characteristics and to avoid introducing new stress concentration points. With this arc-lip configuration, the shell geometry more closely resembled the earlier flexure-free designs than the full-rim geometry of Iteration 5a.

Using the actuation-force test rig and procedure from Section 3, this arc-lip configuration was measured at snap-through forces of 6.7 N, 7.3 N, and 7.8 N across triplicate trials, giving an average of 7.27 N (Table 4.1), against an FEA prediction of 6.36 N (14.3% absolute error). This represents a reduction of approximately 42% relative to Iteration 5a's 12.57 N, and is the configuration referred to going forward as Iteration 5b. However, it still did not meet the desired actuation threshold for magnetic triggering. To reduce the force further while maintaining the same shell characteristics, a depth-tuning procedure was developed and applied to the Iteration 5b geometry, as described in the next subsection.

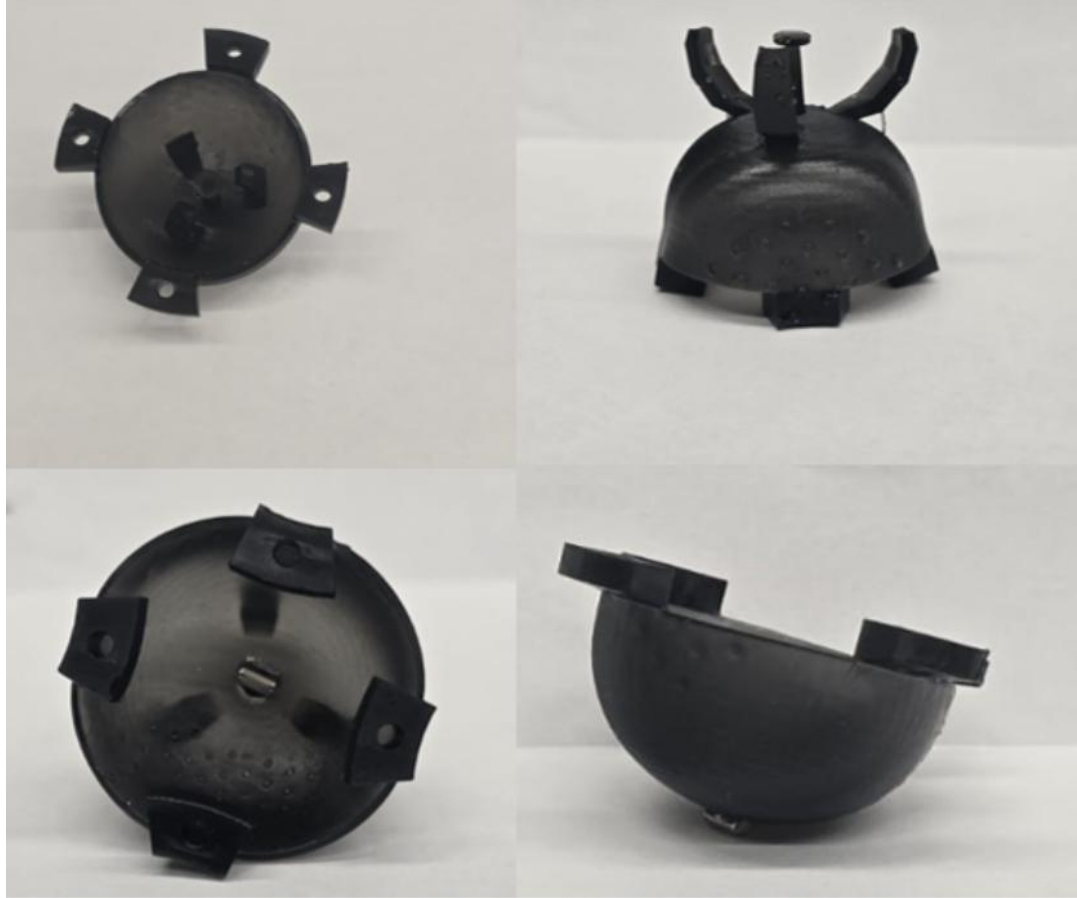


Figure 4.13: Iteration 5b: Top Left (top view with arc sections to hold magnets), Top Right (front view of open gripper), Bottom Left (bottom view of open state), Bottom Right (front view of closed state).

4.5.3 Depth Tuning from Iteration 5b to Iteration 6

To systematically reduce the snap-through actuation force without compromising the shell's intrinsic bistable characteristics, a depth-tuning methodology was developed and applied to the Iteration 5b geometry. As an initial step, the magnet-retaining arc sections were temporarily removed from the assembly, as illustrated in Figure 4.14, to isolate the shell's mechanical response from any geometric perturbations introduced by peripheral features. This allowed the

depth parameter to be evaluated independently as the sole design variable.



Figure 4.14: Depth-tuning procedure applied to Iteration 5b to produce Iteration 6.

The tuning procedure involved incrementally and uniformly reducing the shell's height in discrete steps of a few millimeters each. Following each reduction, the modified shell was fabricated and evaluated for its ability to maintain both bistable states, as shown in Figure 4.15. This iterative process continued until a depth was identified at which a meaningful reduction in snap-through force was achieved while the shell continued to exhibit stable bistable behavior. The convergence criterion was defined as the minimum depth at which both equilibrium states remained self-sustaining under manual perturbation in the absence of external constraints.



Figure 4.15: Checking the open and closed states of the depth-tuned shell derived from Iteration 5b.

An important consideration in this process was preserving the global shell geometry during depth modification. To this end, all dimensional changes were implemented via feature-level operations in SolidWorks rather than at the sketch level. This approach ensured that the underlying shell profile, including curvature and thickness distribution parameters t_1 and t_2 , remained unchanged, and that only the depth parameter was systematically varied, as reflected in the consistent cross-sectional dimensions shown in Figures 4.13 and 4.16. Upon reaching the optimized depth, the magnet-retaining arc sections from Iteration 5b were reintegrated into the design, producing the final form of Iteration 6.

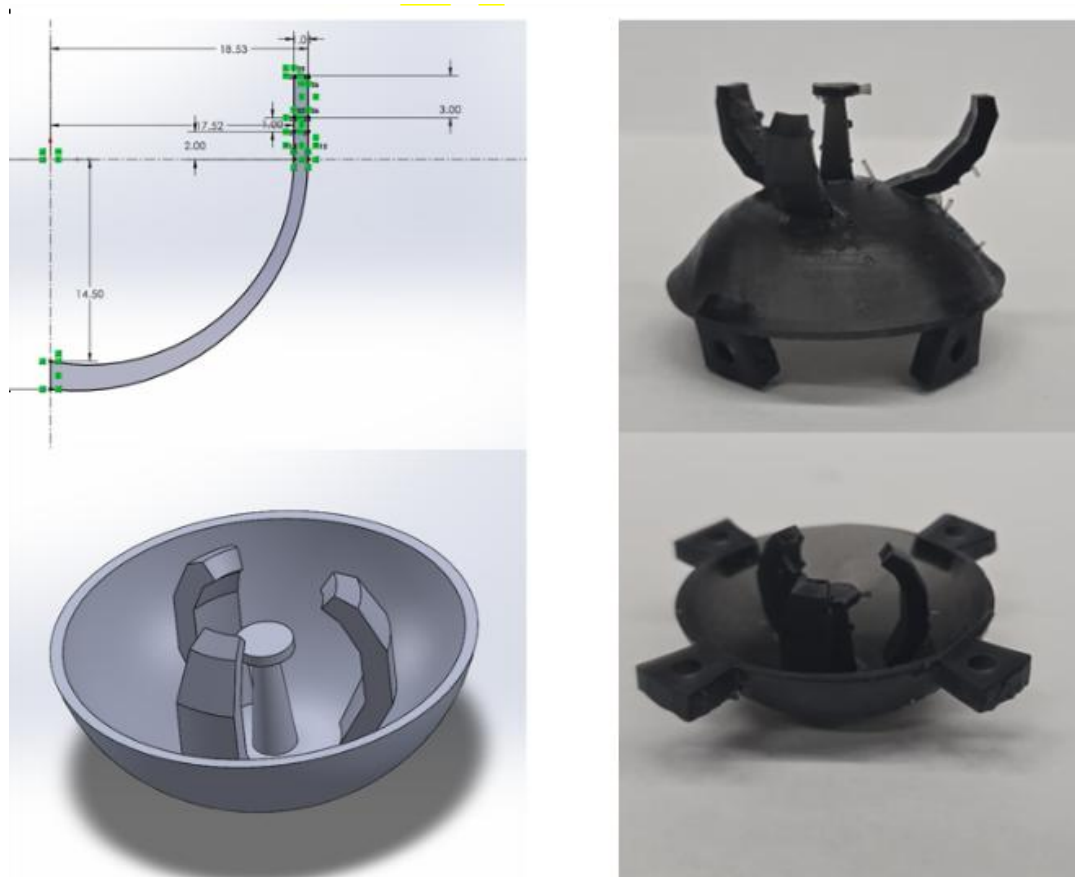


Figure 4.16: Iteration 6: Top Left (shell cross-section), Top Right (closed state), Bottom Left (gripper in SolidWorks), Bottom Right (open state).

4.6 Iteration 6: Composite Assembly and FEA Deviation

Iteration 6 represents the depth-tuned evolution of Iteration 5b, retaining the four-arc magnet-retention geometry while adopting the reduced shell depth identified through the tuning process. For this iteration, the optimized shell was printed in F80 resin without the contact pad or gripper prongs, as shown in Figure 4.17. The prongs and contact pad were then test-fitted and

bonded separately onto the bare shell, using UV-curable resin glue rather than cyanoacrylate, which did not adhere reliably to the F80 surface. The gripper prongs and contact pad for this assembly were fabricated from PLA, resulting in a composite device that combines a compliant resin shell with comparatively rigid PLA gripping features.



Figure 4.17: Iteration 6 shell prior to bonding, with gripper prongs and contact pad unglued.

To quantify actuation performance, the assembled device was placed in its open equilibrium state atop the actuation-force test rig described in Section 3, and triplicate compressive-load trials were carried out until snap-through was observed in each case. The measured snap-through forces were 4.0 N, 3.5 N, and 3.8 N, giving an average of 3.77 N (Table 4.1), compared against an FEA-predicted value of 6.00 N (37.2% absolute error—the largest experiment-to-FEA deviation of any iteration in this study). This represents a reduction of

approximately 48% relative to Iteration 5b's experimental average of 7.27 N, and a cumulative reduction of roughly 74% from the Iteration 2 baseline of 14.23 N.

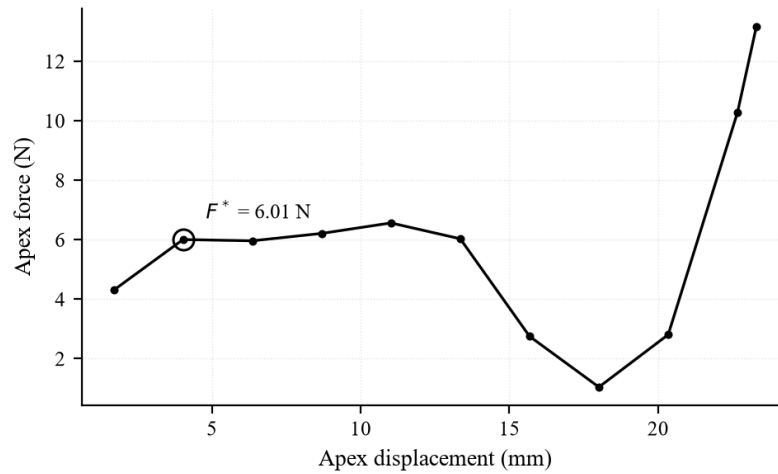


Figure 4.18: Force vs displacement for Iteration 6.

The comparatively large FEA error for Iteration 6 is best understood by considering the device as a composite assembly rather than a single-material shell. The FEA model treats the shell as a standalone, homogeneous Neo-Hookean part, whereas the physical device tested here consists of the F80 resin shell bonded to much stiffer PLA gripper prongs and a PLA contact pad. In earlier single-material iterations, some of the applied force was absorbed as compliant deformation in the prongs and pad themselves, effectively cushioning the load before it reached the shell. Once those features were switched to rigid PLA, the cushioning largely disappeared, and nearly all of the applied compressive force was transferred directly into the shell rather than being partially dissipated by gripper compression. As a result, the shell snapped at a lower externally measured force than the shell-only FEA prediction, which

had no mechanism to represent energy lost in previously softer gripper elements. Future FEA work should therefore model the prongs and contact pad as separate, stiffer bodies bonded to the shell to capture this load-transfer behavior better.

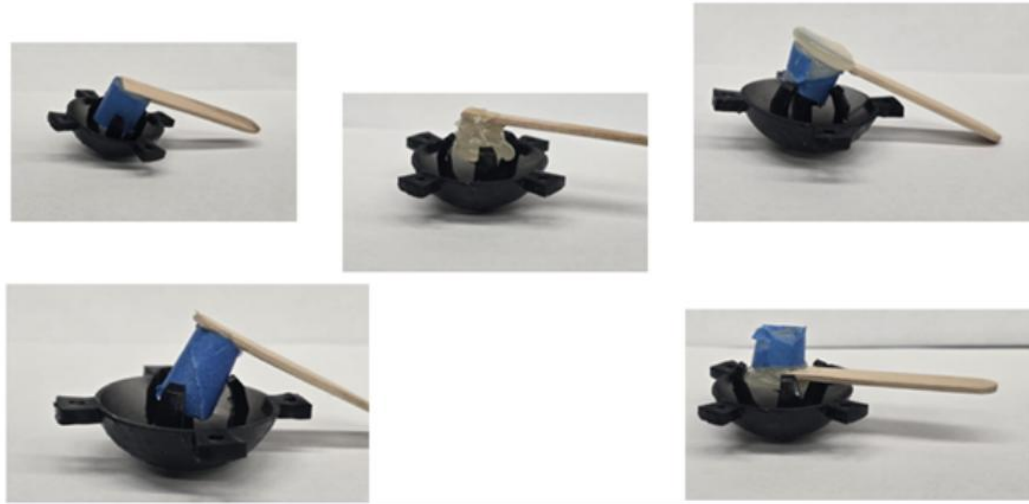


Figure 4.19: Testing Iteration 6 with all test samples (phantoms).

4.6.1 Gripper Design and Performance

Having successfully replicated the shell design in F80 resin, the subsequent phase of this work focused on finalizing the gripper geometry and evaluating its performance across a set of standardized test samples. Four candidate gripper designs were developed, as illustrated in Figures 4.20 and 4.21, each featuring distinct contact geometries intended to accommodate varying tissue surface profiles.

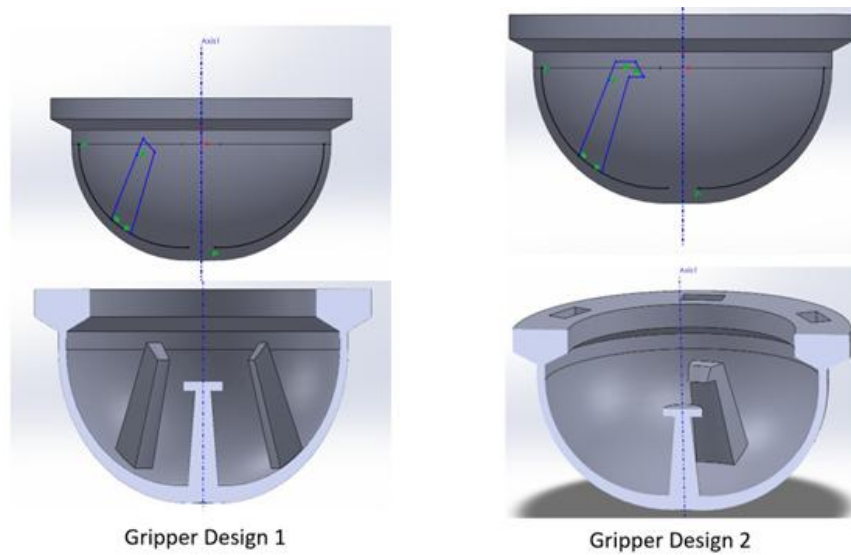


Figure 4.20: Gripper Designs 1 and 2.

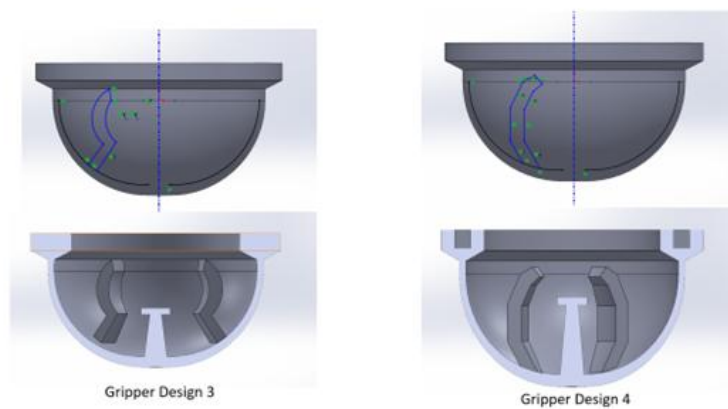


Figure 4.21: Gripper Designs 3 and 4.

Selection of the optimal gripper design was contingent upon several evaluation criteria, foremost among them the mechanical interaction between each gripper and a set of geometrically distinct test samples representing clinically relevant surface contours, as shown in Figure 4.22. These test samples, or

phantoms, were made using hot glue. Hot glue was utilized as a quick, low-cost way to approximate the general size, shape, and semi-compliant feel of the target tissue geometries, rather than as clinically validated tissue surrogates. One phantom was designed to replicate the irregular surface topology characteristic of tumor tissue by randomly pouring hot glue to create irregular surfaces. Others represent conically increasing and decreasing profiles, and a cylindrical surface.

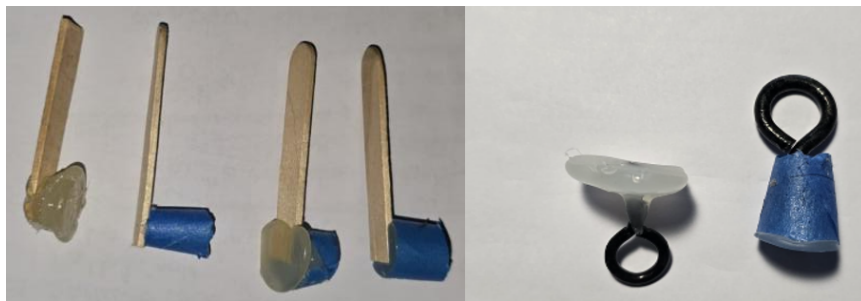


Figure 4.22: Test samples (phantoms) used for gripper evaluation.

In addition to surface interaction, each design was assessed based on the angular clearance between the gripper tips and the test sample surface in the open state, as shown in Figure 4.23. Gripper Design 1 was eliminated early because it failed to achieve adequate gripping contact with most test samples. Gripper Designs 2 and 3 demonstrated satisfactory performance with several geometries. However, each failed to engage at least one phantom (particularly the cylindrical test sample) and exhibited excessive angular deviation between the gripper tips and the sample surface in the open configuration—a characteristic that would compromise reliable actuation in practice. Gripper Design 4 satisfied all prescribed evaluation criteria across the tested phantoms and was advanced for further quantitative testing.

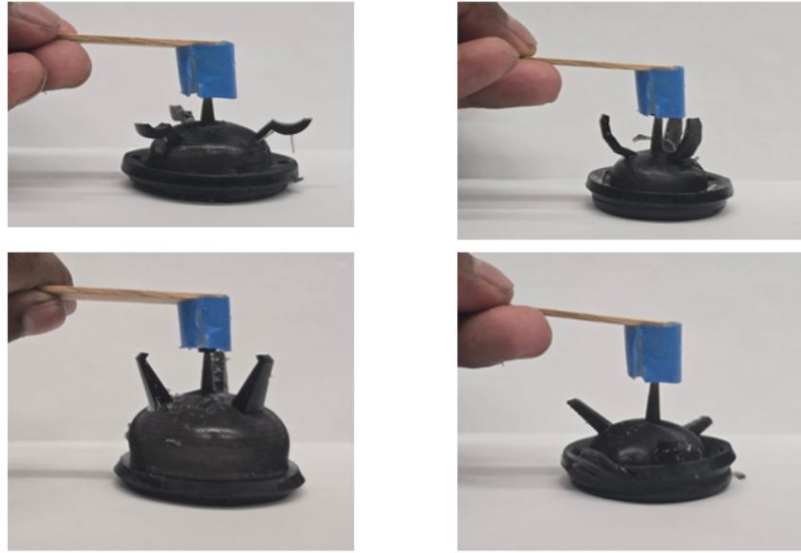


Figure 4.23: Evaluating gripper designs with the test samples.

To quantitatively validate gripper performance, a gravitational load test was conducted for each design. A slotted weight holder was suspended from each gripper, and weights were incrementally added until the gripper could no longer support the load. The mass at failure was recorded and converted to an equivalent force in Newtons via $F = mg$, with $g = 9.81 \text{ m/s}^2$. The resulting grip force values are presented in Table 4.2.

Design	Trial 1 (N)	Trial 2 (N)	Trial 3 (N)	Average (N)
Gripper 1	0.000	0.000	0.000	0.000
Gripper 2	0.098	0.000	0.000	0.033
Gripper 3	0.098	0.000	0.098	0.065
Gripper 4	0.392	0.294	0.392	0.360

Table 4.2: Measured gripping forces for each gripper design, converted from suspended mass.

These quantitative results corroborate the qualitative assessment, affirming Gripper Design 4 as the most capable configuration. The observed trend can

be attributed to differences in contact geometry: Gripper Design 1 exhibited the lowest grip force due to minimal surface contact area and insufficient engagement. Gripper Design 2 demonstrated improved load retention relative to Design 1, yet remained limited by a small number of contact points. Gripper Design 3 achieved a higher grip force than Designs 1 and 2 but was still significantly lower than Design 4; its larger angular separation in the open state would also require greater actuation force to close, making it less suitable for the intended magnetic actuation mechanism. Gripper Design 4 provided the optimal balance of grip strength, contact geometry, and actuation ability, and was therefore selected as the final design for subsequent iterations.

The performance of each gripper design across the primary phantom geometries is summarized in Table 4.3.

Design	Cylindrical Phantom	Descending Cone Phantom (Smaller base)	Ascending Cone Phantom (Larger base)	Polyp Phantom
Gripper 1	No	No	Yes	No
Gripper 2	Yes	No	No	Yes
Gripper 3	No	Yes	Yes	Yes
Gripper 4	Yes	Yes	Yes	Yes

Table 4.3: Qualitative performance of gripper designs across different phantom geometries. “Yes” indicates successful engagement and retention under manual testing.

4.7 Iteration 7

Iteration 7 takes the depth-tuned Iteration 6 shell as its starting geometry and removes material directly from the shell body itself. Specifically, four-quadrant

sections were cut out of the shell surface, following the same fourfold symmetry as the arc magnet-retention sections, to reduce the amount of material that must be elastically deformed during snap-through and, in turn, lower the force required to trigger it. The size, position, and boundary profile of each removed quadrant were iteratively optimized in SolidWorks to preserve sufficient shell structural continuity to maintain reliable bistable behavior. Removing too much material at any one location risked collapsing the shell into a single stable state or exceeding the manufacturing limits of the available resin printer. The F80 resin shell/PLA gripper-and-contact-pad composite construction and UV-resin bonding process followed those described in Iteration 6.



Figure 4.24: Iteration 7 shell geometry with quadrant sections removed from the Iteration 6 baseline.

To characterize actuation performance, the assembled Iteration 7 device was placed in its open equilibrium state atop the actuation-force test rig, and the force-versus-displacement response was recorded through to snap-through, as shown in Figure 4.25. The experimentally measured snap-through forces were 1.9 N, 2.0 N, and 1.9 N, giving an average of 1.93 N (Table 4.1). The FEA model predicts a snap-through point at $F^* = 1.67$ N, and the experimental curve exhibits the same double-hump shape observed in Iteration 6—with an

initial rise, a shallow plateau, and a defined actuation-force peak followed by a steep post-snap climb as the shell moves into its second stable state.

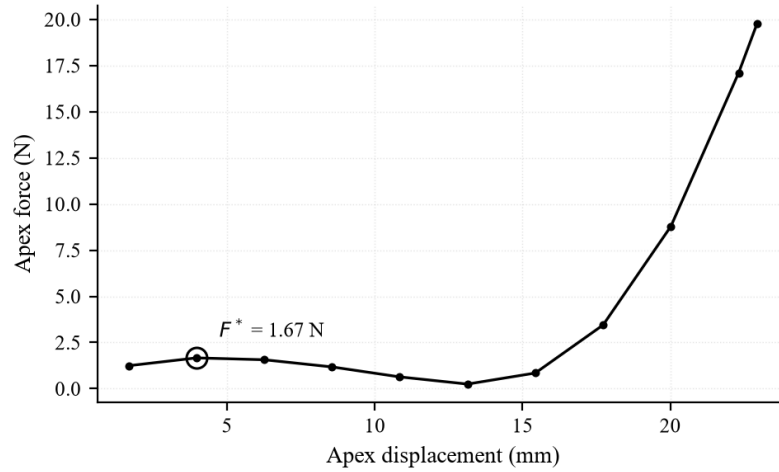


Figure 4.25: Iteration 7 force–displacement curve, showing experimental snap-through behavior and FEA-predicted snap-through point at $F^* = 1.67$ N.

Gripping strength was quantified using the same suspended-load procedure applied to earlier gripper designs, in which incremental mass was added until the gripper released. Iteration 7’s gripper failed at a suspended load of 70 g, which converts to a gripping force of approximately 0.69 N. This is roughly double the grip force measured for Iteration 6’s configuration, and the improvement can be attributed to two main factors: first, the adjusted number and placement of gripper arms allowed more uniform contact around the test sample; second, the thin strip regions introduced by material removal act as additional flexures, enabling more compliant deformation of the gripper tips and increasing the effective contact normal force at the gripping interface.

4.7.1 Reasons for FEA–Experiment Deviation

As with Iteration 6, the experimental snap-through force for Iteration 7 came in below the FEA prediction, and the underlying reason appears to be the same composite-assembly effect. When the full gripper-and-shell assembly was tested rather than the shell in isolation, it became apparent that the earlier, more compliant gripper geometries had quietly absorbed part of the applied compression as their own elastic deformation before that load reached the shell. Once the prongs and contact pad were effectively rigid PLA components glued directly onto the shell, that cushioning effect disappeared, and essentially all of the applied force was transmitted straight into the shell instead of being partially dissipated in gripper compression.

Because the shell is far more compliant than the PLA features bonded to it, this direct load path causes the shell to snap through at a lower externally measured force than the single-material FEA model would predict, since that model lacked a mechanism to represent the energy otherwise lost in compressing a softer gripper. This reinforces the conclusion reached in Iteration 6: modeling the shell as a standalone part, without explicitly capturing the rigid-body coupling introduced by the bonded PLA gripper and contact pad, systematically overpredicts the force needed to trigger snap-through in the finished device. Future modeling work should represent the full composite assembly, including material interfaces and contact mechanics, to improve prediction accuracy.

4.7.2 Scalability and Future Miniaturization

A central question motivating this thesis is whether the actuation-force gap between the bistable shell (on the order of a few Newtons in its current size) and the electromagnetic coil platform’s maximum output force (on the order of 1.34 N in this device configuration, based on the MagnetoSuture platform [13]) can be closed through geometric scaling alone, without a fundamental redesign of the shell concept. To investigate this, snap-through force was characterized at three geometric scales relative to the baseline device: 0.5×, 0.75×, and 1.0×, with the results shown in Figure 4.26.

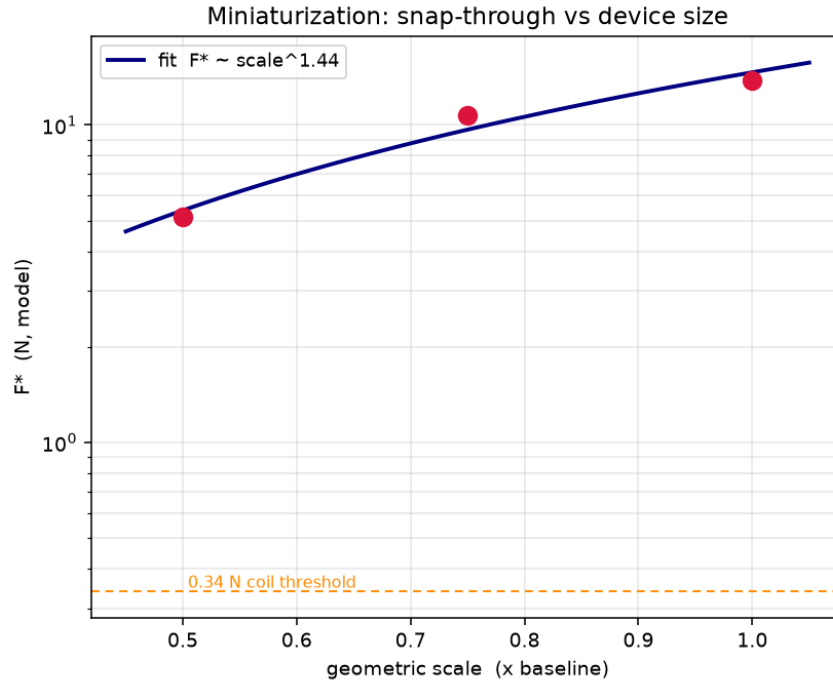


Figure 4.26: Snap-through force versus geometric scale, showing a power-law fit $F^* \sim \text{scale}^{1.44}$ and the 1.34 N coil-platform threshold.

The measured snap-through forces follow a clean power-law trend with

scale, fit as $F^* \sim \text{scale}^{1.44}$, rather than the simple linear relationship that naive compliant-mechanism scaling arguments might suggest for proportionally scaled thickness and curvature. This somewhat steeper-than-linear exponent suggests that, in addition to the geometric scaling of the shell's cross-section, second-order effects, such as changes in relative wall thickness, curvature-to-thickness ratio, or fabrication-driven deviations at smaller scales, also contribute to how the snap-through force falls off as the device shrinks. Extrapolating the fitted trend toward the 1.34 N coil-platform threshold indicates that further, but achievable, miniaturization beyond the 0.5 $\times$ scale tested here should be sufficient to bring the required actuation force within reach of the existing electromagnetic platform, without needing a new shell topology.

This scaling result directly supports the miniaturization strategy proposed at the end of Chapter 2 and reinforces that the actuation deficit identified in this work is primarily a geometric, engineering-scale problem rather than a fundamental limitation of the bistable shell concept itself.

Chapter 5

Conclusion, Discussion and Future Scope

5.1 Conclusion & Discussion

This thesis described the conceptual design and prototype realization of a tetherless, magnetically actuated bistable gripper for fluoroscopy-guided minimally invasive procedures. The device couples a compliant bistable shell with passive gripper prongs to decouple tissue engagement and traction from continuous actuation, allowing it to grip and hold tissue with essentially no steady-state energy input. Iterative design, fabrication in both elastomeric and flexible resin materials, and experimental testing demonstrated repeatable snap-through between open and closed states and tissue-holding forces on representative phantom geometries.

The finite element analysis (FEA) framework developed in this work provides a quantitative basis for design optimization, relating shell thickness, curvature, and geometric scale to snap-through force and stress distribution in

ways that are impractical to obtain through experimentation alone. Miniaturization, shell stiffness, and available magnetic force pull in opposite directions, a central trade-off in this design space that became clear when comparing the measured actuation requirements with magnetic force estimates from the external coil platform. The current prototypes are still larger than the ultimate clinical target and require more actuation force than the platform can currently provide. Nevertheless, they confirm that the combination of bistable compliant shells and remote magnetic actuation is a viable path to tetherless gripping under fluoroscopic guidance.

Collectively, these results suggest that tetherless bistable grippers can overcome core limitations of conventional tethered polypectomy and traction tools—namely constrained dexterity, dependence on direct line-of-sight, and unwanted force transmission through rigid shafts and cables—by enabling localized, remotely actuated tissue engagement even within tortuous anatomy.

5.1.1 Key Findings and Contributions

The main technical contributions of this dissertation are:

- **Design constraints and dimensional requirements.** Reported specifications for anatomical and robotic colonoscopy were used to formulate design constraints and dimensional requirements for a colon-targeted, tetherless gripping device.
- **Conceptual device architecture.** A conceptual device architecture was developed that combines a bistable compliant shell, passive gripper

prongs, embedded permanent magnets, and an external electromagnetic coil platform for orientation and state switching.

- **Iterative fabrication workflow.** An iterative fabrication workflow using silicone elastomers and F80 flexible resin was implemented to tune shell geometry, stiffness, and manufacturability. This resulted in prototypes with repeatable snap-through and measurable holding forces across multiple phantom geometries.
- **Complementary FEA pipeline.** A complementary FEA pipeline was developed that predicts snap-through forces across a geometric design space, reveals stress-concentration and failure modes, and elucidates a power-law relationship between geometric scale and required actuation force,

$$F^* \sim \text{scale}^{1.44}.$$

- **Magnetic actuation integration.** Preliminary integration with the Mag-netoSuture coil system and magnet orientation strategies demonstrated remote control of device state and orientation, and outlined requirements for future closed-loop magnetic control and actuation.

These contributions provide a foundation for a next generation of tetherless gripping tools that can be miniaturized, multiplexed, and functionally specialized for minimally invasive surgery.

5.1.2 Discussion of Limitations

Several limitations remain that should inform the interpretation of these findings and suggest future directions:

- **FEA–experiment deviation.** Differences between the predicted and measured snap-through forces are attributed to simplifications in the material models, uncertainty in the boundary conditions, and fabrication imperfections such as variable shell thickness and residual stress. Closing this gap will likely involve more accurate material characterization (e.g., nonlinear viscoelastic models), tighter process control, and better registration between simulated and as-fabricated geometries.
- **Actuation margin.** The magnetic force available from the current coil platform is not yet sufficient to independently actuate highly miniaturized shells. Design changes that make the structure stiffer or more geometrically robust tend to raise the force threshold for snap-through, further reducing the margin between what the shell requires and what the platform can provide.
- **Scalability.** The scalability study in this thesis showed that the snap-through force obeys a power-law relationship with geometric scale, $F^* \sim \text{scale}^{1.44}$. This is a steeper relationship than the linear trend predicted by simple compliant-mechanism theory, indicating that further miniaturization reduces the required actuation force more rapidly than expected. Extrapolation of this trend suggests that scales beyond the $0.5 \times$ baseline tested here should bring the required snap-through force

within reach of the current coil platform's 1.34 N ceiling. However, this relationship was characterized only at three data points ($0.5\times$, $0.75\times$, and $1.0\times$) and has yet to be validated at the sub-millimeter feature sizes relevant to true capsule-scale deployment, where print resolution and material fatigue behavior become limiting factors in their own right. This scaling relationship further informs miniaturization.

- **Clinical integration.** The prototypes described in this work are still at benchtop scale and were tested in simplified phantom setups. To move toward a clinically deployable device, biocompatible materials and coatings, sterilizable and reproducible fabrication processes, and rigorous testing in anatomically realistic phantoms and preclinical models will be necessary.

These limitations frame this thesis as an enabling step toward a clinically viable device, rather than a finished clinical solution.

5.2 Future Scope

5.2.1 Multimaterial Composite Fabrication

The work of Khatua et al. [41] demonstrated that fiber orientation is a dominant factor governing the structural performance of composite materials. In particular, randomly distributed (isotropic) fiber arrangements provided only marginal improvements in bulk stiffness compared with fibers aligned along the principal stress directions.

Inspired by these findings, the configurations shown in Figure 5.1 were

explored to investigate whether sparsely distributed F80 inclusions and linearly aligned F80 strips could provide a manufacturing strategy capable of achieving an effective balance between flexibility and stiffness.

Although this approach remained at the preliminary investigation stage and was not pursued further due to time constraints, it represents a promising direction for the proposed device architecture. Such a strategy would enable selective tailoring of materials across different subcomponents, with the gripper fabricated from higher-stiffness materials. At the same time, the bistable shell is manufactured from more compliant materials to preserve flexibility.

Furthermore, linearly oriented reinforcing fibers could be embedded within the shell structure to constrain deformation directions or to selectively reinforce the bistable mechanism locally.

Beyond stiffness tailoring, a further extension of the gripper prongs themselves is worth considering: adding a small cutting or shearing feature directly onto the prong tips. As explored in a companion investigation of a tetherless magnetic-cutting variant of this device, tissue engagement and cutting can be mechanically decoupled by pairing the passive bistable gripping shell with an externally driven rotating magnetic field that produces shear-based cutting at the prong interface once the tissue is secured. In that architecture, the cutting energy is encoded in the rotational frequency of the external field rather than in the raw magnitude of the applied force, sidestepping a key limitation of existing tetherless magnetic scissors designs that depend directly on field strength for cutting power. Retrofitting a small blade or shearing edge onto the existing gripper prongs, rather than introducing an entirely separate



Figure 5.1: Conceptual configurations using F80 inclusions and aligned reinforcing strips for stiffness tailoring.

5.2.2 Future Scope of FEA

geometry — parametric solid

depth d

tip radius

clasp

clasp

iteration rules & optimization

iteration	rule
1	build
2	invert
3	probe
4	compare
5	optimize

hooks

iteration rules & optimization

While the parametrization of the gripper-arm topology and placement was successful, Figure 5.4 shows a time-step simulation of Iteration 1 interacting

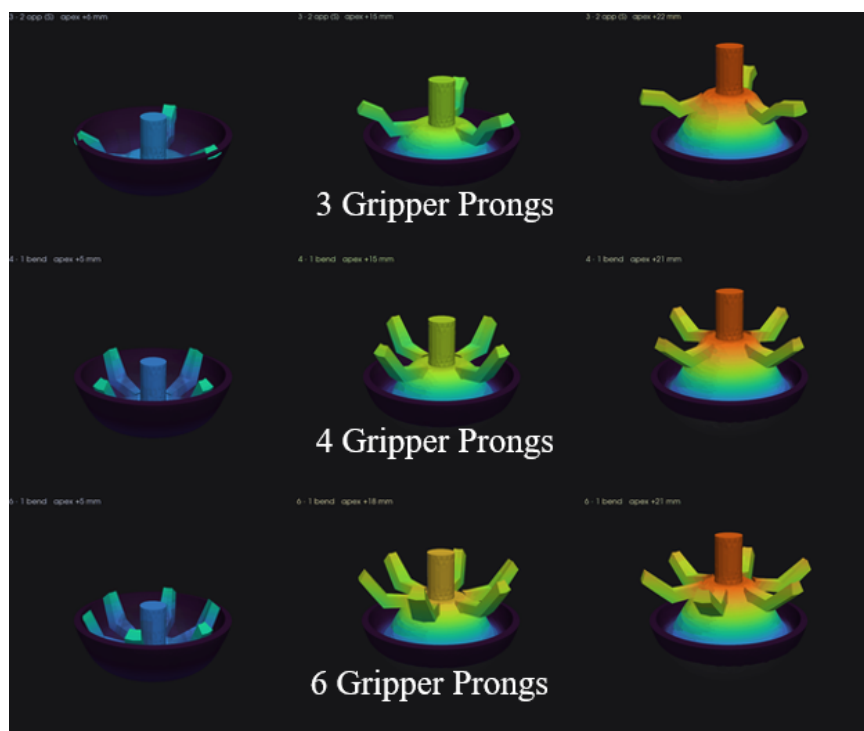


Figure 5.3: Parametrized multiple-arm gripper models.

with a bulb-on-rod phantom, which approximates a polyp on a stem in the rectum. The main drawback of the simulation was that contact-force modeling was not initialized properly: forces were displayed on the bulb rather than on the gripper arms. The simulation also did not clearly show the deformation and stresses experienced by the arms. Addressing these issues would be a valuable future direction and would help bring FEA predictions closer to experimental results.

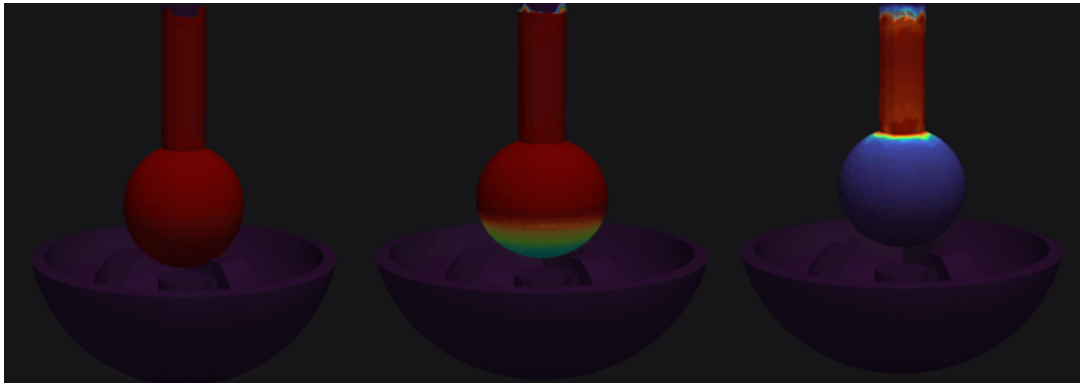


Figure 5.4: Time-based simulation of the gripper interacting with a polyp-approximation phantom.

5.2.3 Actuation Platform and Control

Although the present actuation system was not sufficiently strong to reliably activate the larger prototype, future work may use the VORIACS platform [14] to provide greater magnetic-field strength and articulation, with the software architecture adapted from Chen et al. [42].

Future clinical experiments could use fluoroscopy to track robots. Chen et al. [43] demonstrated that magnetic-field modeling based on the Zernike model, combined with nonlinear model predictive control, can enable accurate

robot control despite the low frame rate and measurement noise inherent to fluoroscopic imaging.

5.2.4 Possible Extension to Gripper Prongs/Devices for Colon Application

The bistable gripping platform described in this thesis is primarily conceived as a mechanically efficient, tetherless device for atraumatic tissue engagement and stabilization during colonic polypectomy [40]. While the current design focuses on passive compliant gripping and magnetic actuation, the gripper prongs also provide a natural substrate for integrating additional diagnostic and therapeutic functions tailored to colonic applications. In particular, the inner surfaces and tips of the prongs can be used for independent local sensing, hemostasis, thermal treatment, and drug-delivery modules that act synergistically with the underlying bistable shell mechanics.

5.2.4.1 Application 1: Real-time tactile sensing and haptic feedback for minimally invasive biopsy

Recent advances in bio-inspired tactile-sensing capsule endoscopy have demonstrated that it is feasible to equip an untethered endoscopic pill with a deformable sensing surface or force-sensing elements capable of detecting subtle changes in the gastrointestinal wall beyond what is visible to onboard cameras [44, 45]. In such systems, the capsule shell is instrumented with a distributed tactile interface (for example, soft biomimetic structures or force-sensing cantilevers) whose deformation under contact with the intestinal wall is transduced into measurable signals, enabling detection and characterization

of submucosal lesions during locomotion through the gut [44, 45]. By analyzing the spatiotemporal patterns of these tactile signals as the capsule traverses the lumen, the device can infer local changes in wall stiffness, lump geometry, and tissue loading associated with buried or early-stage tumors that do not produce obvious surface discoloration or gross morphological changes.

A similar tactile-sensing modality could be miniaturized and embedded along the inner faces of the bistable gripper prongs, effectively turning each prong into a localized tactile probe that interrogates the mechanical environment of a target polyp or surrounding mucosa. In this configuration, small arrays of force-sensitive structures (for example, micro-cantilevers or compliant tactile ridges) on the prong surfaces would deform as the gripper is closed, and the resulting signal patterns could be processed to estimate local contact forces, indentation depth, and stiffness distribution across the gripped tissue [36, 40]. During polyp engagement or minimally invasive biopsy, these measurements would allow the operator to keep gripping forces within atraumatic ranges while simultaneously providing quantitative information about lesion consistency as a proxy for pathology.

Although current tactile capsule platforms are primarily designed for autonomous screening and diagnosis, the same class of tactile signals can be mapped to a haptic interface on the operator side, for instance, by driving an external vibrotactile or force-feedback device in proportion to measured contact forces or reconstructed stiffness profiles [44]. In the context of the tetherless bistable gripper, such a closed-loop configuration would enable

remote palpation-like interaction. As the prongs tent the polyp and surrounding mucosa, the evolving tactile signal field could be streamed wirelessly and rendered as haptic feedback at the surgeon console, providing an additional modality beyond fluoroscopic imaging for assessing tissue consistency and biopsy adequacy. This capability is particularly valuable when optical information is limited or absent, such as in fluoroscopy-guided interventions where tactile feedback is otherwise completely decoupled from the surgical site.

5.2.4.2 Application 2: Liquid-metal microcoil prongs for localized cauterization

Room-temperature gallium-based liquid metals have emerged as promising materials for soft biomedical electronics due to their high electrical and thermal conductivity, intrinsic deformability, and biocompatibility [46, 47]. Recent work highlights their use in stretchable heaters, neural interfaces, and thermotherapy systems, where patterned liquid-metal traces embedded in elastomeric substrates can generate controllable Joule heating when driven by modest currents [46]. Translating these concepts to the bistable gripper, thin serpentine liquid-metal microcoils could be integrated within each prong, with the coil cross-section positioned close to the tissue-contact surface and electrically insulated by a thin biocompatible polymer layer. When energized by an external RF or DC power source coupled through the magnetic actuation platform, these microcoils would locally elevate tissue temperature along the gripping interface, enabling controlled cauterization of resection margins following polypectomy. Because the gripper is already mechanically

anchored to the polyp base, the thermal energy could be applied in a spatially confined manner, potentially reducing collateral damage relative to traditional monopolar cautery delivered through a long tethered snare.

5.2.4.3 Application 3: Theragripper-inspired drug delivery

Ghosh et al. [48] introduced submillimetre “theragrippers” that autonomously close onto colonic mucosa when a thermoresponsive wax layer softens at body temperature, penetrating superficially and releasing ketorolac over many hours, thereby extending drug half-life from roughly 2 to about 12 hours in rat colon. Instead of deploying thousands of free microdevices, similar multiclawed, drug-loaded microstructures could be fabricated directly on or near the inner surfaces of the gripper prongs, so that once the gripper snaps onto the polyp base, a subset of these units latches onto the resection bed and elutes analgesic or anti-inflammatory agents locally. Because the theragripper fabrication flow is wafer-based and compatible with thin-film processes, integrating an array of such microdeposits onto a macro-scale bistable shell is conceptually feasible. It would turn the gripper–tissue interface into a controlled, tetherless drug-delivery zone [48].

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