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Los Angeles

Design, Modeling and Control of Flexible Structures in Robotics Applications

A dissertation submitted in partial satisfaction
of the requirements for the degree
Doctor of Philosophy in Mechanical Engineering

by

Zhuonan Hao

2025

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ABSTRACT OF THE DISSERTATION

Design, Modeling and Control of Flexible Structures in Robotics Applications

by

Zhuonan Hao

Doctor of Philosophy in Mechanical Engineering

University of California, Los Angeles, 2025

Professor Mohammad Khalid Jawed, Chair

Flexible structures play a pivotal role in enabling novel forms of robotic locomotion, manipulation, and medical intervention. This dissertation investigates the design, modeling, and control of flexible structures across increasing geometric complexity—from rods, to rings, to shells—with an emphasis on their integration into robotic systems. First, we examine rod-like geometries inspired by bacterial flagella, where single and bi-flagellated robots leverage helical rod rotation to achieve propulsion in low Reynolds number environments. Second, we present pneumatic ring actuators that generate peristaltic motions for grasping and transporting irregular or soft objects, demonstrating robust manipulation without the need for complex control strategies. Finally, we extend the framework to pneumatic membrane shells for hemostasis applications, coupling soft structures with robotic arms to achieve precise deployment and pressure regulation on bleeding sites. Across all three case studies, we develop physics-based and data-driven models, integrate proprioceptive and external sensing for feedback, and design control strategies that link structural mechanics with task-specific robotic performance. Together, these contributions establish a unified perspective on how flexible structures—ranging from slender rods to curved shells—can be harnessed to expand the functionality, adaptability, and safety of next-generation robotic systems.

The dissertation of Zhuonan Hao is approved.

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CHAPTER 1

Introduction

1.1 Motivation

Soft robotics has rapidly emerged as a paradigm shift in how machines are designed, modeled, and deployed. Unlike traditional rigid robots, soft robotic systems are constructed from compliant materials that can deform continuously, adapt to unstructured environments, and interact safely with humans. This intrinsic compliance opens new opportunities across domains ranging from manufacturing and environmental exploration to biomedical interventions. However, the very properties that make soft robots attractive also make them challenging to design and control: their dynamics are governed by nonlinear elasticity, large deformations, and strong coupling between geometry, material response, and environment.

Among the diverse forms of soft structures, three archetypes appear repeatedly: **rods**, **rings**, and **shells**. These canonical geometries capture the fundamental ways slender and thin structures deform and transmit forces. Rods describe elongated bodies that bend, twist, and buckle; rings represent closed-loop structures that can actuate cyclically to transport matter; and shells generalize to two-dimensional surfaces that deform in complex ways under pressure. Each archetype is not only of theoretical significance in mechanics, but also underpins practical designs in robotics and medicine. This dissertation develops and unifies methods for modeling, controlling, and applying rod-, ring-, and shell-based soft robotic systems, with particular emphasis on biomedical applications where compliant interaction is critical.

1.2 Research Themes

This dissertation builds upon three central themes, corresponding to the three archetypes:

Rods: Modeling and Control of Slender Structures

In Chapter 3, we investigate the mechanics and control of soft robotic rods. Using the discrete elastic rod (DER) model, we formulate a simulation-driven control strategy that inverts the traditional paradigm: instead of predicting deformation from forces, we identify the forces required to achieve a desired configuration. This inversion enables predictive control of slender structures, offering insights into locomotion, navigation, and shape regulation for catheter-like devices and bio-inspired robots. The framework provides a foundation for robust, real-time control of slender elastic systems in both engineering and biomedical contexts.

Rings: Peristaltic Locomotion and Transport

Chapter 4 focuses on ring-like actuators arranged in modular sequences to produce peristaltic locomotion. Inspired by biological transport mechanisms, the design enables grabbing and moving delicate objects through synchronized inflation and deflation of multiple rings. Finite element simulations and experimental testing demonstrate how geometry and synchronization govern efficiency and adaptability. Beyond robotics, such mechanisms hold promise in underwater exploration and minimally invasive medical transport systems, where handling irregular or fragile objects is essential.

Shells: Soft Membranes for Hemostasis

In Chapter 5, we extend our framework to thin shell structures, exemplified by a pneumatically actuated membrane designed for robotic hemostasis. Unlike rods and rings, shells provide distributed surface contact, which is crucial for safe interaction with soft biological tissue. By embedding visuo-tactile and ultrasound sensing within the membrane, we achieve

multi-modal feedback that enables closed-loop compression of blood vessels. Experimental validation on phantom models and preliminary in vivo studies highlight how shell-based designs can reduce variability, improve safety, and pave the way for intelligent clinical devices.

1.3 Unifying Perspective

Although rods, rings, and shells differ geometrically, they are united by three key principles:

1. **Nonlinear mechanics:** All undergo large, nonlinear deformations that demand beyond-linear elasticity models.
2. **Coupling of geometry and function:** Their performance arises from geometric nonlinearities—whether in bending stiffness of rods, cyclic deformation of rings, or pressure-driven curvature of shells.
3. **Opportunities for biomedical application:** Each archetype enables new forms of safe, adaptive interaction with biological systems.

This unifying view allows the dissertation to bridge fundamental mechanics with application-driven design. By systematically studying rods, rings, and shells, the work builds a coherent framework for advancing soft robotics at the intersection of mechanics, control, and biomedicine.

1.4 Contributions

The main contributions of this dissertation are:

- A simulation-driven control framework for rods based on the discrete elastic rod model, enabling predictive control of slender soft actuators.
- The design and validation of ring-based peristaltic actuators for transport and manipulation of delicate objects.

- A shell-based robotic hemostasis platform integrating visuo-tactile and ultrasound sensing for adaptive, image-guided vascular compression.
- A unified perspective connecting rod, ring, and shell mechanics as archetypal geometries for biomedical soft robotics.

1.5 Organization of the Dissertation

The dissertation is structured as follows:

- Chapter 2 reviews background material on soft robotics, elastic rod and shell theory, sensing integration, and clinical challenges in hemostasis.
- Chapter 3 presents the modeling and control of rod-like soft actuators using discrete elastic rod theory.
- Chapter 4 introduces the design and analysis of ring-based peristaltic actuators for transport tasks.
- Chapter 5 describes the shell-based robotic hemostasis platform, integrating multi-modal sensing and validating performance in phantom and in vivo settings.
- Chapter 6 concludes with a summary of findings and future research directions.

CHAPTER 2

Background and Related Work

2.1 Foundations of Soft Robotics

Soft robotics has emerged as a paradigm that departs from traditional rigid-link mechanisms, enabling robots to interact with unstructured environments, delicate biological tissues, and uncertain operating conditions with a degree of safety and adaptability previously unattainable [1], [2]. Inspired by biological systems such as octopus arms, worms, and membranes, soft robots exploit material compliance to achieve shape morphing, distributed actuation, and energy-efficient locomotion. Their ability to conform to contact surfaces reduces stress concentrations and failure risks, particularly in biomedical contexts [3]–[5].

From an engineering standpoint, soft robots rely on advanced materials (silicones, hydrogels, elastomers), diverse actuation mechanisms (pneumatic, hydraulic, shape-memory alloy, dielectric elastomer), and robust sensing strategies. Hyperelastic constitutive modeling is essential for predicting material behavior under large deformations, with frameworks such as Mooney–Rivlin or Ogden models providing analytical foundations [6]. Nonlinear behaviors of shells and membranes further require computational strategies such as finite element analysis or reduced-order models [7].

Computational advances extend beyond materials. The discrete elastic rods (DER) framework allows simulation of slender filaments, capturing bending, twisting, and stretching [8]. Fluid–structure interaction models, including resistive force theory and regularized Stokeslets, enable prediction of motion in viscous environments [9]. More recently, robust rod contact handling and implicit methods extend these models to dynamic robotic scenarios [10].

Given this diversity, soft robotics research can be organized into three canonical archetypes: (1) rods, inspired by slender-body locomotion such as bacterial flagella; (2) rings, modeled after peristaltic transport and tubular organs; and (3) shells, inspired by compliant surfaces and membranes. These archetypes capture recurring biological principles and provide a unifying framework for robotic innovation.

2.2 Rods and Flagellar Locomotion

2.2.1 Biological Flagella and Cilia

Flagella and cilia are ubiquitous in microbiology, serving as primary locomotion mechanisms in viscous, low-Reynolds-number environments [11]–[13]. The bacterial flagellar motor rotates a helical filament, converting torque into propulsion [14], [15]. Classical work by Gray and Hancock [16] and Berg [17] established that reciprocal strokes are ineffective at low Reynolds number (the “scallop theorem”), necessitating non-reciprocal waveforms.

Flagella also exhibit complex instabilities. Buckling instabilities facilitate bacterial tumbling [18]–[20], while polymorphic transformations enable switching between propulsion and reorientation [21], [22]. Bundling and unbundling of multiple flagella regulate swimming modes [23]–[25]. These biological insights provide a foundation for bio-inspired robotics.

2.2.2 Robotic Flagella Analogs

Scaling these principles to macroscopic robots has yielded significant progress. Early macroscopic models validated hydrodynamic predictions of helical propulsion [26]. Robotic swimmers with multiple flexible filaments demonstrated bundling behavior consistent with bacterial locomotion [27], [28]. Magnetically actuated swimmers at the microscale [29], [30] and flexible filament robots at the macroscale [31], [32] highlight applications ranging from targeted drug delivery to environmental monitoring.

2.2.3 Computational Rod Mechanics

Mechanically, rods represent slender, deformable structures governed by Kirchhoff’s rod theory or DER formulations [8]. Computational studies extend to instabilities, knots, and dynamic contacts [10]. The coupling of rod mechanics with hydrodynamics via resistive force theory, slender-body theory, or Stokeslet-based methods provides predictive models for propulsion efficiency and trajectory control [9]. Applications extend to robotic navigation in confined spaces, steerable catheters, and flexible manipulators.

2.3 Rings and Peristaltic Locomotion

2.3.1 Biological Peristalsis

Peristalsis refers to sequential muscle contractions that transport material through tubular organs such as the esophagus and intestines [33]–[35]. Peristaltic transport underlies digestion, excretion, and circulatory processes, with autorhythmic activity patterns characterized by contractile waves [36]. Biological scaling studies show how body geometry affects crawling and bolus transport [37], [38]. Regenerative medicine highlights the difficulty of replicating peristaltic function in engineered tissues [39].

2.3.2 Peristaltic Robots

Roboticians have long drawn inspiration from earthworms and gastrointestinal peristalsis. Early peristaltic robots employed SMA actuation [40], pneumatic artificial muscles [41], or servomotors [42]. Capsule endoscopes leverage peristaltic locomotion for minimally invasive diagnostics [43], [44]. Recent designs employ modular soft actuators mimicking esophageal swallowing [45] or intestinal transport [46].

Control strategies optimize locomotion efficiency. Kandhari et al. [47] studied trade-offs between velocity and cost of transport, while Peng et al. [48] demonstrated adaptive control in inchworm-inspired devices. Applications extend to microgravity waste transport

systems [49], [50], demonstrating the robustness of peristaltic principles.

2.3.3 Actuation and Design Trends

Modern peristaltic actuators employ bellows-driven pneumatics for lightweight, controllable actuation [51]. Soft peristaltic robots such as Meshworm employ antagonistic NiTi coils for crawling [52]. Reviews highlight design diversification across materials and actuation methods [53]. Rings thus embody rhythmic locomotion and transport, bridging biological mechanisms with engineered devices.

2.4 Shells and Membrane Structures

2.4.1 Hyperelastic Membrane Mechanics

Membranes and shells represent the surface-based archetype of soft robotics. Their hyperelastic mechanics enable conformal contact, wide-area force distribution, and morphing. Toroidal actuators [7], toroidal hydrostats [54], and torus-inspired graspers [55] illustrate how geometry governs function. Constitutive modeling using Mooney–Rivlin frameworks enables prediction of inflation, wrinkling, and failure [6].

2.4.2 Biomedical and Robotic Applications

Biomedical soft shells serve in wound compression, compression therapy, and hemostasis [3], [56]. Epidermal electronics demonstrate integration of sensing with compliant substrates [4]. In robotics, shell-like structures enable underwater grasping [5], deployable structures, and programmable shape morphing **guseinov2020programming**. These examples highlight the versatility of shells for distributed contact tasks.

2.4.3 Comparative Perspective

Compared to rods and rings, shells uniquely provide two-dimensional surface actuation. This enables applications where wide-area, adaptive contact is required—whether for safe tissue compression, underwater manipulation, or deployable robotic systems.

2.5 Comparative Literature Summary

Table 2.1 summarizes representative biological inspirations, robotic implementations, and key references across rods, rings, and shells.

Table 2.1: Comparison of rods, rings, and shells in biology and robotics.

Archetype	Biological Examples	Robotic Prototype	References
Rods	Bacterial flagella, cilia, sperm tails	Magnetic microrobots, flexible filament swimmers, DER/DDG-based manipulators	[11], [16], [27], [57]
Rings	Esophageal bolus transport, intestinal peristalsis, earthworm crawling	Capsule endoscopes, peristaltic crawling robots, space waste transfer systems	[33], [37], [45], [52]
Shells	Biological membranes, toroidal hydrostats, pressure-distributing skin interfaces	Hyperelastic toroidal actuators, soft robotic graspers, biomedical compression devices	[3], [7], [54], [55]

2.6 Research Pipeline

The research presented in this dissertation follows a structured pipeline, as illustrated in Fig. 4.2. The process begins with the identification of both the *structure* to be studied and its intended *application*. These two elements provide the conceptual foundation from which a draft design is formulated. The draft stage establishes the essential geometry, actuation scheme, and material considerations, enabling subsequent exploration of feasibility.

From this stage, the design evolves along two parallel but interconnected paths: the **sim-**

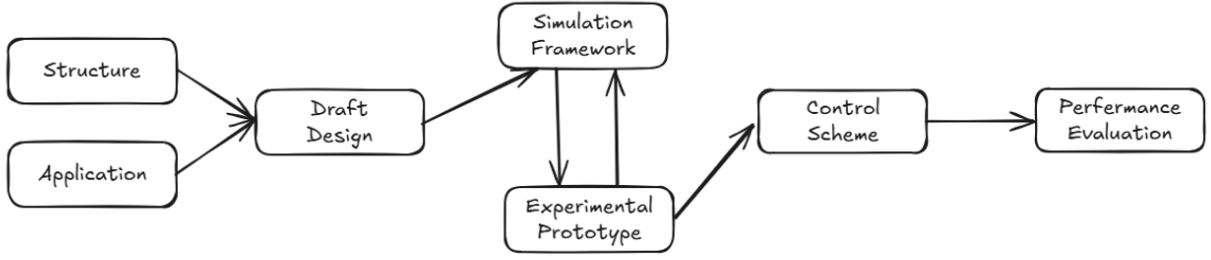


Figure 2.1: Research pipeline showing the iterative process from conceptual design to performance evaluation.

ulation framework and the **experimental prototype**. The simulation framework plays a critical role in accelerating the design process. By providing low-cost, rapid, and highly adaptable testing environments, simulations enable broad parameter sweeps and sensitivity analyses that would be impractical through fabrication alone. Importantly, the insights from simulations not only guide structural optimization but also provide intuition for control scheme formulation. In parallel, physical prototyping serves as a ground-truth reference, validating the assumptions and predictions of the computational models while uncovering practical challenges in fabrication and operation.

The dual insights from simulation and experiment converge to inform the **control scheme**. In this stage, actuation strategies, feedback mechanisms, and coordination of modules are developed and refined. Control formulation is not treated as an afterthought but rather as an iterative design driver, ensuring that the mechanical and sensing systems remain compatible with real-time operation.

Finally, the pipeline culminates in **performance evaluation**, where the integrated system is tested under representative conditions. Metrics of efficiency, robustness, and adaptability are quantified, and feedback from this stage may cycle back into earlier phases of the pipeline, further improving the design.

In summary, this pipeline is inherently iterative and integrative: by combining the predictive power of simulations with the realism of experiments, and by continuously feeding these insights into control and evaluation, the research process produces systems that are both theoretically grounded and practically robust.

2.7 Summary

This chapter has reviewed the conceptual and technical foundations of soft robotics and organized the field through three canonical archetypes: rods, rings, and shells. Each of these categories emerges from strong biological inspiration and reflects a fundamental class of mechanics, actuation, and control strategies.

Rods capture the physics of slender filaments such as flagella and cilia, which serve as natural solutions for propulsion at microscopic scales. Their study has led to significant advances in discrete elastic rod theory, fluid–structure interaction modeling, and robotic systems capable of reproducing swimming, steering, and navigation in highly viscous or constrained environments. Beyond locomotion, rod-like structures form the basis for steerable catheters, continuum manipulators, and surgical instruments, demonstrating their broad relevance across engineering and medicine.

Rings highlight the principle of rhythmic peristalsis, a biological mechanism that underpins digestion, circulation, and tubular transport. Robotic analogues of peristalsis showcase how sequential actuation of soft segments can generate reliable transport even in confined or complex environments. These designs span from earthworm-inspired crawlers to capsule endoscopes and peristaltic waste transfer systems for microgravity. The versatility of peristaltic architectures lies in their ability to combine anchoring and propulsion in a single mechanism, offering efficient locomotion strategies with strong potential for modular, scalable systems.

Shells extend the discussion to surface-level, hyperelastic structures that can deliver distributed contact forces across wide areas. Unlike rods or rings, which emphasize line-based or sequential interactions, shells enable conformal coverage and programmable morphing. Biological membranes, toroidal hydrostats, and compliant skin interfaces motivate robotic designs such as inflatable graspers, toroidal actuators, and biomedical compression devices. These systems illustrate how geometric and material nonlinearities can be leveraged for adaptive pressure distribution, conformal sensing, and multifunctional actuation.

Taken together, these three archetypes—rods, rings, and shells—offer complementary perspectives on how soft robotics can harness material compliance and bioinspiration. They

also provide methodological building blocks for integrated systems: rods capture slender-body dynamics and instabilities; rings demonstrate coordinated, distributed actuation for transport; and shells showcase hyperelastic membranes for safe surface interaction. Importantly, each archetype highlights different scales of interaction, from microscale propulsion in fluids to macroscale conformal contact with tissues.

By synthesizing insights across biology, mechanics, and robotics, this chapter establishes a framework for the remainder of this dissertation. The following chapters expand upon these archetypes through detailed investigations of rod-based propulsion, ring-based peristaltic transport, and shell-based compression mechanisms. Together, these studies aim to bridge fundamental mechanics with applied robotic design, ultimately advancing the broader vision of safe, adaptive, and bio-inspired soft robotic systems.

CHAPTER 3

Rod Structures for Flagellated Propulsion

3.1 Overview

The study of rod-like structures offers a fundamental gateway into understanding how geometry, flexibility, and fluid interaction combine to generate propulsion in soft robotic systems. Among these, helical flagella stand as one of the most iconic and effective natural designs for locomotion in viscous environments. From the nanoscale machinery of bacteria to bio-inspired robotic prototypes, soft helical filaments illustrate how slender, flexible rods can convert rotational input into thrust, exploit nonlinear instabilities, and adapt to complex surroundings.

This chapter focuses on developing both experimental and computational frameworks for investigating the propulsion, stability, and control of rod-like soft flagella. By scaling up flagella into centimeter-scale robotic analogs, we gain precise control over geometric and material properties, while still reproducing the low Reynolds number regime characteristic of bacterial swimming. These macroscopic models allow systematic exploration of thrust generation, directional buckling, and multi-flagellum hydrodynamic interaction—phenomena that are challenging to measure directly at the microscale.

To complement physical experiments, we employ advanced simulation methods based on discrete elastic rods (DER), regularized Stokeslet segments (RSS), and implicit contact modeling (IMC). This combination captures the nonlinear elasticity of soft filaments, long-range hydrodynamic coupling, and filament–filament contact mechanics. Together, experiment and simulation form a unified platform for revealing how flagellar flexibility, rotation direction, spacing, and geometry influence performance.

The outcomes of this work extend beyond biological analogy. By elucidating the interplay of elasticity and hydrodynamics, rod-based studies inform the design of artificial microswimmers, catheter-like continuum robots, and filament-driven biomedical devices. More broadly, the principles uncovered here illustrate a central theme of this dissertation: that flexible rods, though simple in form, exhibit rich mechanics that can be harnessed to expand the capabilities of soft robotic systems.

3.2 Mechanics of Soft Helical Flagella: Thrust, Buckling, and Hydrodynamic Interaction

3.2.1 Background

The propulsion mechanisms of bacteria operating in viscous, low Reynolds number environments have emerged as a rich platform for studying the coupled physics of elasticity and fluid dynamics. Unlike macroscopic swimmers that rely on inertial forces, bacteria achieve locomotion through rotating flagella – slender, flexible filaments that deform and interact with the surrounding fluid in fundamentally nonlinear ways. This insight traces back to the pioneering work of Berg [11], who developed early techniques for tracking bacterial motion, and later visualized flagellar bundling and unbundling using fluorescence microscopy [22].

Classical hydrodynamic models, such as Resistive Force Theory (RFT) [16], approximate local drag forces to capture the motion of flagellated swimmers. Despite its continued use, RFT inherently neglects long-range fluid-mediated interactions – a factor now known to be central to flagellar propulsion. Recent studies have emphasized that flexibility is not merely a passive trait but a functional feature enabling critical behaviors such as buckling, bundling, and periodical reorientation, and other polymorphic transitions – crucial for navigation and steering [13], [18], [57]. These phenomena arise from a combination of geometrically nonlinear deformations and long-range hydrodynamic interactions, both of which play an active role in microscale propulsion.

While numerical simulations and biological observations have highlighted these complex-

ities, direct and systematic experimental measurements remain scarce. Natural flagella offer limited experimental control: tuning geometric parameters such as pitch, radius, or spacing is difficult, and biological variability complicates repeatability. Moreover, the forces involved are minuscule and difficult to isolate, especially in multi-flagella systems where inter-filament contact and synchronization further obscure interpretation.

One of the most notable efforts to measure flagellar force in biological systems was by Darnton et al. [21], who quantified thrust from tethered *Salmonella*. To bypass the limitations of microscale experiments, researchers have developed macroscopic analogs. An early and influential example was a scaled model of flagellar bundling by Kim et al. [26], which demonstrated the utility of tabletop systems in visualizing bacterial hydrodynamics. A major advance came from Rodenborn et al. [23], who used rigid steel-wire helices in viscous fluids to measure force, torque, and drag. Their results showed that RFT fails to predict even single-flagellum behavior, while long-range hydrodynamic models provided accurate agreement – highlighting the inadequacy of local drag approximations in capturing collective effects.

Despite these contributions, the role of filament flexibility – and its resulting instabilities such as buckling – has only recently been investigated [20], [57]. Yet, to date, there exists no experimental study that directly and systematically measures the thrust generated by flexible flagella, especially under conditions of close proximity or coordinated actuation.

In this work, we address this gap by developing a centimeter-scale experimental platform for studying rotating soft helical flagella in high-viscosity fluids. This model system allows precise control over geometric and actuation parameters—such as helix pitch, radius, stiffness, angular speed, and inter-flagellum spacing—while enabling direct measurement of propulsive forces. We complement these experiments with simulations that couple Discrete Elastic Rod (DER) mechanics [8], Regularized Stokeslet Segments (RSS)[9] for long-range hydrodynamics, and an Implicit Contact model (IMC)[10] for physical filament interactions. Unlike simplified models such as resistive force theory (RFT), our framework captures both nonlinear structural instabilities and fluid-mediated coupling. By systematically varying geometric and kinematic inputs, we reveal how geometry, elasticity, and inter-flagellum interactions govern force production. The close agreement between experimental and simu-

lated results underscores the importance of accurately modeling long-range hydrodynamics and structural nonlinearity, providing new insights into elasto-hydrodynamic propulsion and informing the design of artificial microswimmers and medical microrobots.

3.2.2 Physical and Numerical Experiments

3.2.2.1 Macroscopic Experiment Setup

Experiments on the soft flagella were performed in a 43-liter (11.3-gallon) acrylic tank with internal dimensions of $350 \times 350 \times 350$ mm, filled with silicone oil (Clearco PSF-5000cSt Silicone Fluid). The fluid had a measured density of 975 kg/m^3 and dynamic viscosity of $4.465 \text{ kg} \cdot \text{m}^{-1} \cdot \text{s}^{-1}$ at 25°C , approximately 4465 times the viscosity of water. Viscosity was verified using a cone-and-plate viscometer (USS-DVT6), and the oil’s temperature was monitored using a thermometer (Elitech RC4). The manufacturer-reported temperature coefficient of viscosity is $1.0 \times 10^{-6} \text{ kg} \cdot \text{m}^{-1} \cdot \text{s}^{-1}/^\circ\text{C}$ and thermal expansion coefficient is $9.4 \times 10^{-4}/^\circ\text{C}$.

Each flagellum was driven by a DC motor (Pololu Micro Metal Gearmotor HPCB 6V with an Encoder), enclosed in a 3D-printed housing. Axial thrust was measured using a SparkFun TAL221 load cell (1 N full scale), amplified by an HX711 module. The data was transmitted through an Arduino to a PC for recording. Motor control was implemented through a SparkFun L298N motor driver. For each trial, thrust was measured under both clockwise and counterclockwise rotation across ten discrete speeds ranging from 4 to 100 rpm. Repeatability between CW and CCW measurements was within 2%. The Reynolds number was kept below 0.1 in all tests, calculated using $Re = \rho\omega Rr_0/\mu$, where ρ is the fluid density, ω is angular speed, R is the helical radius, r_0 is the cross-sectional radius, and μ is the dynamic viscosity.

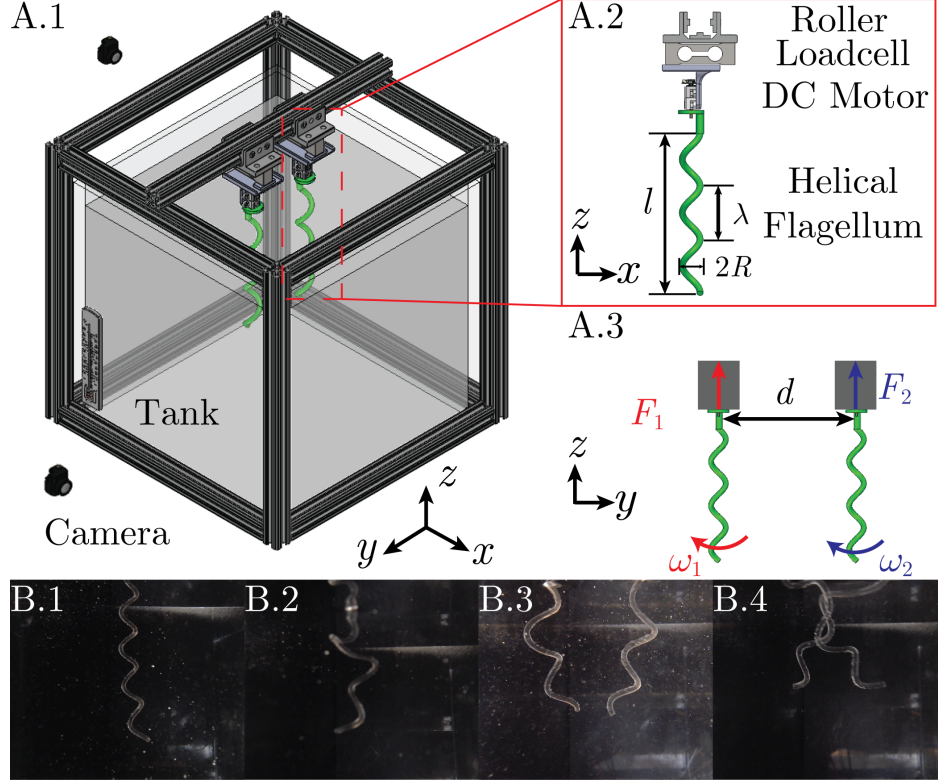


Figure 3.1: (A) Schematic diagram of the experimental setup capable of measuring the thrust generated by two or more flagella, crucial for studying their interaction. Key components include the tank, load cell, and helical flagellum driven by a DC motor. (B) Fabrication process of the soft flagellum using a mixture of iron powder and VPS material, molded with a silicone iron mold, and supported by a PVC tube and syringe for precise shaping.

3.2.2.2 Fabrication of Bionic Flagellum

Soft helical flagella were fabricated using Vinyl Polysiloxane (VPS) elastomer (Zhermack Elite Double series: 8, 16, 22, 32). To adjust the density of the material, iron powder (density $7,800 \text{ kg/m}^3$) was added to the VPS base and catalyst mixture. The resulting composite was injected into pre-shaped PVC tubes (McMaster-Carr Masterkleer Soft PVC Tubing), which were used as molds. After curing, the tubes were slit open to extract the soft flagella. The resulting helices had a cross-sectional radius of 6 mm, helical radius of 8.89 mm, pitch of 76 mm, and axial length of 196.2 mm. The fabricated samples exhibited Young's moduli ranging from 0.18 to 1.30 MPa and densities between 1020 and 1180 kg/m^3 , with an assumed Poisson's ratio of 0.5. For comparison, rigid flagella with identical geometry were 3D printed using polylactic acid (PLA), with a Young's modulus of 4.11 GPa and density of

1250 kg/m³.

3.2.2.3 Numerical Simulations Using DER Framework

Numerical simulations were performed using the Discrete Elastic Rods (DER) method [8], a framework grounded in discrete differential geometry that captures geometrically nonlinear bending, twisting, and stretching of slender structures. Each flagellum was modeled as a piecewise linear centerline coupled with a per-segment orthonormal material frame. Hydrodynamic interactions were incorporated using the Regularized Stokeslet Segment (RSS) method [9], allowing long-range fluid-mediated coupling under low Reynolds number conditions. Physical contact between flagella was handled using an implicit contact (IMC) model [10].

All input parameters for geometry, material properties, rotation speed, and flagella spacing were taken directly from measurements without fitting. Simulations were run in parallel with experiments, and quantitative comparisons were made to validate the interplay between flexibility, hydrodynamics, and contact. Additional implementation details and sensitivity analysis are provided in the SI Appendix.

3.2.2.4 Normalization Scheme

The helical flagellum is connected to the head at one extremity, where it is rotated counterclockwise with a prescribed rotation speed ω . Two flagella are spaced with a specific distance d . Hereafter, we normalize the spacing distance by the helix radius, R , such that the normalized distance is $\bar{d} = d/R$, where the overbar symbol $\bar{\cdot}$ denotes the normalized variables. Likewise, the geometrical parameters that describes the flagellum structure include $\bar{\lambda} = \lambda/R$ and $\bar{L} = L/R$. The input for the bi-flagellated system is angular velocity ω as a function of time. The propulsive thrust F is the z component of $\mathbf{f}_p^{\text{tail}}$, with the normalization as $\bar{F} = F/(\mu\omega RL)$, where μ denote the viscosity of fluid. This dimensionless representation allows for generality across length scales in interpreting our findings.

Understanding the dynamics of rotating helical structures in viscous fluids requires dis-

secting how geometry, material properties, and environmental interactions collectively influence propulsion and stability. This section presents a series of experimental and computational studies designed to validate the thrust-generation model, quantify direction-dependent buckling thresholds, and examine the role of flexibility in single-body and multi-body interaction scenarios. Together, these analyses reveal how design and control strategies must be adapted when transitioning from rigid to soft helical systems, and when considering the behavior of multiple interacting units.

3.2.3 Results and Discussion

3.2.3.1 Validation of Thrust Measurement System

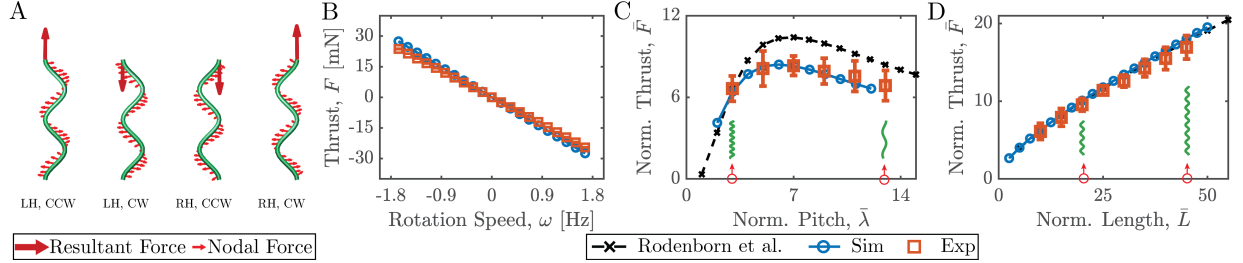


Figure 3.2: Thrust generation of rotating helical flagellum. (A) Direction of net thrust force generated by rotating helices depends on both handedness (left-handed (LH) or right-handed (RH)) and rotation direction (clockwise (CW) or counter-clockwise (CCW)). Red arrows indicate nodal forces along the filament, and the bold arrow indicates the resulting net force. (B) Quantitative comparison of thrust F as a function of rotation rate ω , showing excellent agreement between simulation (Sim) and experimental (Exp) results. Inset: thrust fluctuations over time for a specific rotation rate. (C) Normalized thrust $\bar{F}$ as a function of normalized pitch $\bar{\lambda}$, demonstrating geometric influence on propulsion. (D) Thrust increases linearly with normalized length $\bar{L}$, consistent with previous work by Rodenborn et al[23].

We validate our simulation framework by exploring how the thrust force generated by rotating helical structures depends on geometric and kinematic parameters. Figure 3.2(A) illustrates how the combination of helix handedness and rotation direction determines the direction of the net thrust force. Left-handed helices rotating CCW and right-handed helices rotating CW generate upward thrust, while the opposite combinations generate downward thrust, as a result of nodal force summation along the helical body. Figure 3.2(B) shows that the thrust force magnitude varies linearly with rotation rate, with simulation results closely

matching experimental measurements across the full range of angular velocities. The inset confirms temporal stability of the force during steady rotation. [Figure 3.2\(C\)](#) investigates how normalized pitch affects propulsion efficiency; both experimental and simulation data reveal an optimal pitch range that maximizes thrust. Finally, [Figure 3.2\(D\)](#) demonstrates a nearly linear relationship between normalized helical length and thrust force, consistent with earlier experimental studies, indicating that longer helices proportionally increase force output.

3.2.3.2 Directional Buckling Instability

[Figure 3.3](#) characterizes the buckling thresholds of soft helical filaments under rotation, highlighting the directional dependence of instability onset. When the filament is rotated counter-clockwise (denoted as ω_b^+) or clockwise (ω_b^-), it experiences distinct buckling thresholds due to the geometric asymmetry inherent in the helical structure.

[Figure 3.3\(A\)](#) presents the critical buckling rotation rates as a function of normalized pitch $\bar{\lambda}$. Both experimental (triangles for ω_b^+ and squares for ω_b^-) and simulation results (dashed lines) consistently show that the buckling threshold for CCW rotation (ω_b^+) is significantly higher than that for CW rotation (ω_b^-), and the gap widens with increasing pitch. This asymmetry arises from the non-centrosymmetric geometry of the helix: rotation in one direction tends to compress and stabilize the structure, whereas the opposite direction enhances destabilizing bending or twisting deformations.

[Figure 3.3\(B\)](#) explores how the normalized length $\bar{l}$ influences the buckling thresholds. Both simulation and experimental data again reveal a clear asymmetry between ω_b^+ and ω_b^- , with longer filaments exhibiting lower buckling thresholds overall. Notably, the difference between the two rotational directions becomes more pronounced as length increases, further underscoring the role of geometric asymmetry in shaping stability characteristics. These results emphasize that both the direction of rotation and the filament geometry must be considered when designing soft helical systems for propulsion or fluidic applications.

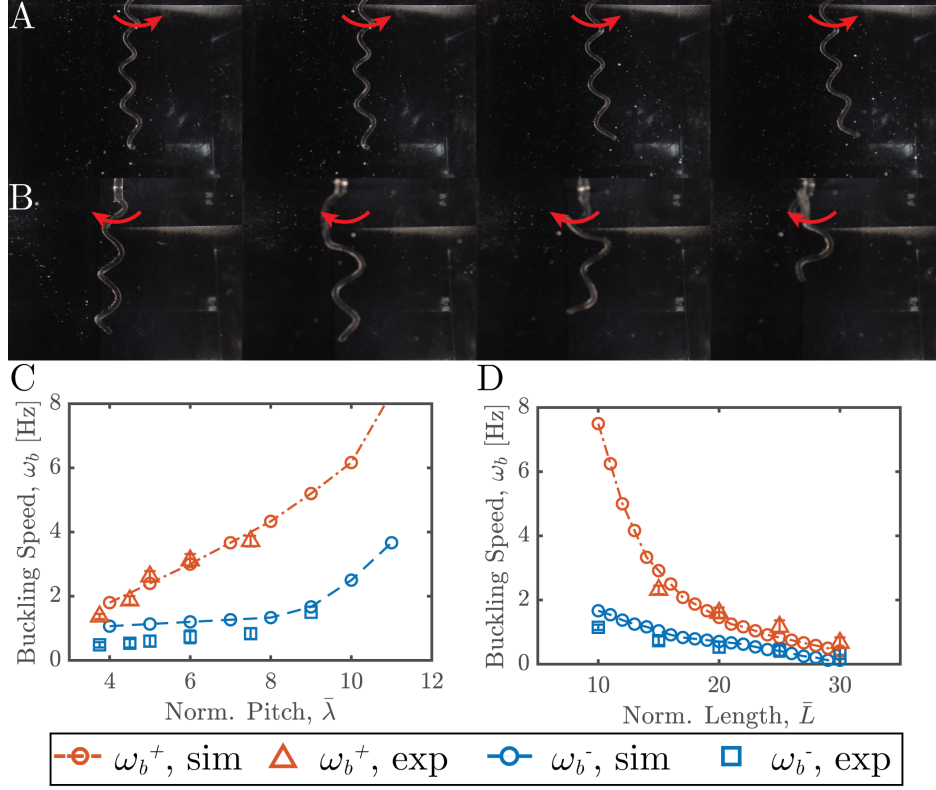


Figure 3.3: Buckling thresholds ω_b for soft helical filaments rotating in opposite directions: counter-clockwise (ω_b^+) and clockwise (ω_b^-). (A) The threshold rotation rate as a function of normalized pitch $\bar{\lambda}$, with both simulation (dashed lines) and experimental (symbols) results shown. The CCW case (ω_b^+ , red) consistently exhibits higher buckling thresholds than the CW case (ω_b^- , blue), and this difference grows with increasing pitch. (B) Buckling thresholds as a function of normalized length $\bar{L}$. As the filament becomes longer, the threshold decreases for both directions, with a more pronounced asymmetry between ω_b^+ and ω_b^- . The observed directional dependence arises from the intrinsic asymmetry of the helical geometry, which causes the filament to respond differently to opposite rotation directions.

3.2.3.3 Multi-body Hydrodynamic Interactions

Beyond the behavior of single filaments, understanding the dynamics of multiple interacting helical structures is crucial for designing collective swimming systems or manipulating fluids in confined environments. This section investigates how hydrodynamic and potential mechanical interactions between neighboring rotating helices influence their individual and collective thrust generation. Figure 3.4 illustrates three distinct interaction scenarios based on the relative rotation directions and separation distance between two helical filaments. In the D0 configuration, a single reference filament rotates in isolation, generating a baseline

thrust. The D1 configuration features two identical helices rotating in the same direction ($\omega_1 > 0$ and $\omega_2 > 0$) with varying separation distance d . The D2 configuration examines two helices rotating in opposite directions ($\omega_1 > 0$ and $\omega_2 < 0$), also with varying d . Figure 3.4(B) quantifies the influence of initial phase difference ($\Delta\phi$) on thrust generation for a specific D1 configuration at a fixed separation. Both simulation and experimental data show that the thrust generated by individual filaments remains largely insensitive to the initial phase difference between them. This suggests that for this specific interaction type and spacing, the primary hydrodynamic effects are largely independent of the relative rotational phasing of the helices. Figure 3.4(C) presents the thrust generated by individual filaments ($F_{\text{individual}}$) as a function of their separation distance d for the D1 and D2 configurations, compared to the D0 baseline. For the D1 case (same rotation direction), individual thrust decreases significantly as the filaments approach each other, especially at smaller distances. This reduction can be attributed to increased hydrodynamic interference and viscous coupling between the co-rotating helices. Conversely, for the D2 case (opposite rotation direction), the individual thrust is substantially reduced compared to both D0 and D1, and it shows a less pronounced dependence on distance, suggesting strong destructive interference from counter-rotating flows. Finally, Figure 3.4(D) shows the total sum of thrust generated by the two filaments (F_{sum}). In the D1 configuration, the total thrust initially increases with decreasing distance from a baseline (twice the D0 thrust, shown by the dashed yellow line), but then declines as the filaments become very close, indicating a complex interplay of beneficial and detrimental hydrodynamic interactions. For the D2 configuration, the summed thrust remains very low across all distances, often near zero or even negative, highlighting the strong cancellation effects when helices rotate in opposite directions. These results underscore the critical role of relative rotation direction and inter-filament spacing in dictating the overall propulsive efficiency of multi-helical systems.

3.2.3.4 Contact Interaction

We investigate the role of structural flexibility and local interactions in determining the propulsive force generated by rotating helical filaments. Figure 3.5(A) presents a control

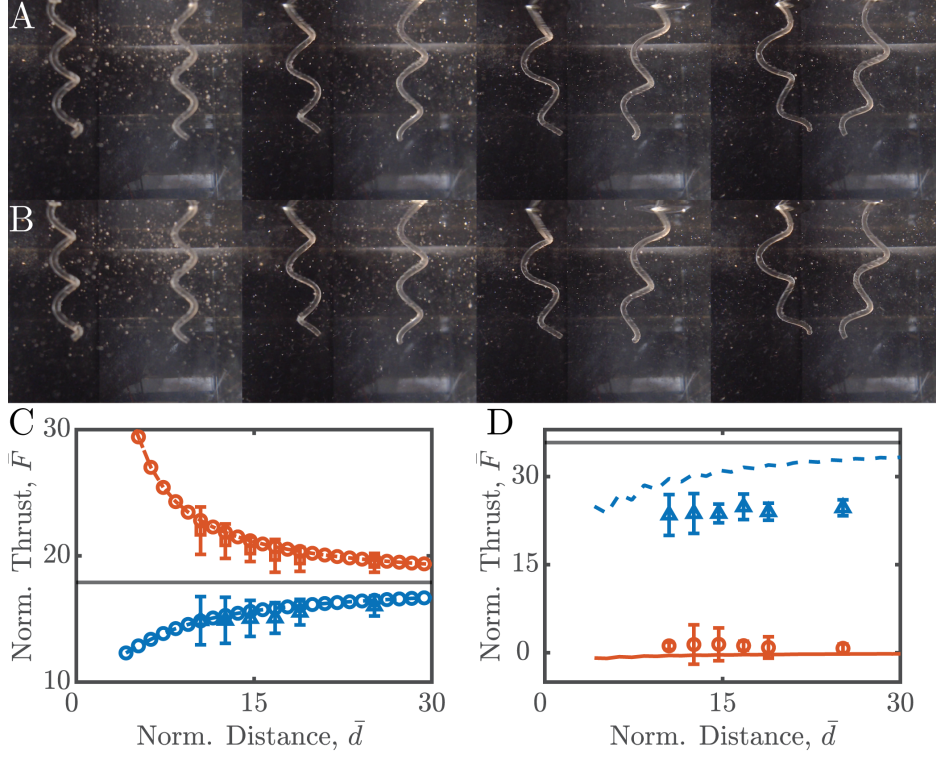


Figure 3.4: Hydrodynamic and Mechanical Interactions in Multi-body Helical Systems. (A) Schematic illustration of three interaction scenarios: D0 (single isolated helix), D1 (two helices rotating in the same direction, $\omega_1 > 0, \omega_2 > 0$), and D2 (two helices rotating in opposite directions, $\omega_1 > 0, \omega_2 < 0$). The parameter d represents the center-to-center separation distance between the helices. (B) Influence of initial phase difference ($\Delta\phi = |\phi_1 - \phi_2|$) on the thrust magnitude for a D1 configuration at a fixed separation, showing insensitivity to relative phasing. Simulation (Sim) and experimental (Exp) results are compared. (C) Individual thrust force ($F_{\text{individual}}$) as a function of separation distance d . For D1, individual thrust decreases significantly at smaller distances, while for D2, thrust is greatly reduced on distance. The D0 baseline (dashed yellow line) represents the thrust of a single isolated helix as a baseline. (D) Summed thrust force (F_{sum}) generated by both filaments as a function of separation distance d . In the D1 configuration, total thrust is less than twice the D0 baseline at intermediate distances, whereas for D2, the summed thrust remains low across all distances, often near zero.

case in which a pair of rigid helical filaments is rotated at a constant rate in a viscous fluid environment. In this configuration, thrust arises purely from fluid-structure interaction and serves as a reference for evaluating more complex cases.

Figure 3.5(B) illustrates the case of a soft helical filament with identical geometric parameters and rotation rate. Due to its compliance, the soft filament undergoes additional deformation under viscous loading, which in turn alters its hydrodynamic profile and reduces

its ability to generate thrust. Furthermore, if the soft filament is in proximity to neighboring filaments, both hydrodynamic and mechanical contact interactions can further diminish net propulsion.

To systematically quantify this effect, [Figure 3.5\(C\)](#) compares the thrust differential $\Delta F = F_{\text{rigid}} - F_{\text{soft}}$ over a range of rotation rates ω . Across the entire range, ΔF remains negative, highlighting a consistent reduction in thrust in the soft filament relative to its rigid counterpart. This difference reflects a combination of fluid-mediated interaction effects (e.g., wake interference) and contact interactions (e.g., collisions), with the latter becoming increasingly significant as structural flexibility increases. Both experimental data and simulation results are shown, confirming the robustness of the observed trends.

Overall, we quantify the extent to which flexibility and interaction effects impair thrust generation in soft rotating systems.

3.3 Bundling and Tumbling in Bacterial-inspired Bi-flagellated Soft Robots for Attitude Adjustment

3.3.1 Background

Bacterial locomotion has long stood as a captivating frontier in the realm of robotics and bio-inspired engineering [\[12\]](#). The design and control of highly maneuverable robotic systems have been informed by the propulsion mechanisms observed in nature, especially those employed by motile bacteria [\[17\]](#). The Scallop theorem implies that traditional propulsion mechanisms, like fins, paddle, and wings, become inefficient in highly viscous environment [\[13\]](#). The microorganisms, like E. coli, often equipped with slender helical appendages protruding from their cells, exhibit remarkable agility in their aquatic environments [\[15\]](#), [\[58\]](#). Drawing inspiration from the natural world’s efficiency and adaptability, roboticists have sought to emulate these bacterial propulsion methods with single [\[30\]](#)–[\[32\]](#) or multiple flagella [\[27\]](#), [\[29\]](#), [\[59\]](#). Multi-flagellated swimmer have mainly two modes of locomotion, i.e., run and tumble, via the intricate interplay of multiple slender flagella [\[14\]](#). However, cur-

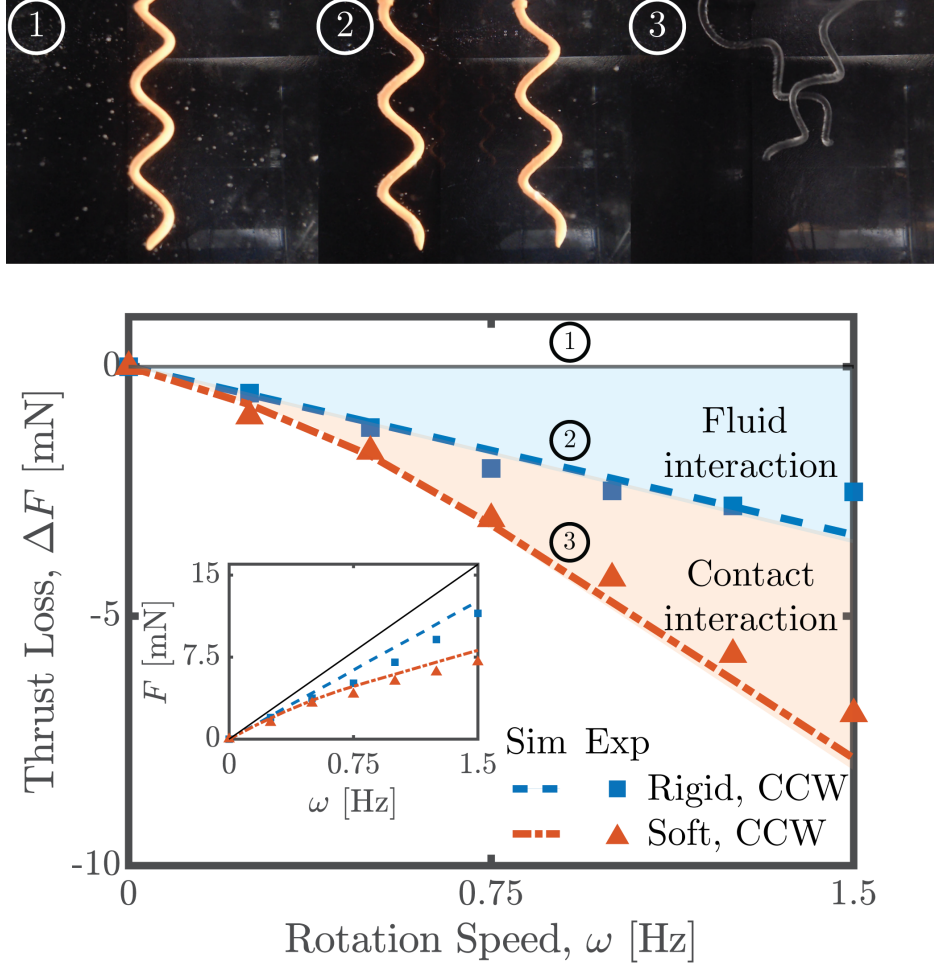


Figure 3.5: Hydrodynamic and contact interaction. (A) Thrust force F generated by a single rigid helical filament (D1, Rigid, CCW) in the absence of physical contact, serving as a baseline. (B) Thrust force produced by a soft filament (D1, Soft, CCW) under the same rotation rate, showing reduced force generation. (C) Comparison of thrust changes ΔF between the rigid and soft cases as a function of rotation rate ω . The net reduction in thrust, indicated by negative ΔF , reflects the influence of both fluid interaction (blue region) and contact interaction (orange region) in the soft system. Simulation and experimental results are shown in both cases.

rent research primarily emphasizes the running ability of multi-flagellated robots, with less attention directed toward the realization of direction-changing mechanisms.

Recognizing this gap, we delve into the realm of attitude adjustment for multi-flagellated robots. The slender structure of these robots exhibits intriguing polymorphic transitions, including buckling, bundling, and tumbling [18], [19], [24], [25], [57], setting them apart from traditional rigid robots. Their soft, deformable flagella provide them with remarkable agility

and the ability to navigate intricate and confined spaces. However, the intricate nonlinear nature of their structure, coupled with the complexities of fluid and contact interactions, underscores the need for in-depth exploration.

Since it is difficult to systematically manipulate the physical parameters of microscopic natural bacteria, we choose an analog robotic system which can - in turn - inform the original natural system that motivated its design. This macroscopic bi-flagellated system is affixed to a ball joint, to facilitate the characterization and quantification of its attitude adjustment capabilities. As visualized in Figure 4.1, the robot is submerged in viscous fluid medium to faithfully emulate the environment bacteria thrive in. This robotic system exhibits the capacity to rotate along different body axes, i.e., yaw and pitch angle, reproducing the bacterial ability. Furthermore, to understand this phenomenon and explore the parameter space governing attitude adjustment, we introduce a comprehensive computational framework. This framework addresses critical aspects such as elasticity, long-range hydrodynamics, and the handling of physical contacts, each of which has demonstrated its efficacy in prior research [10], [27].

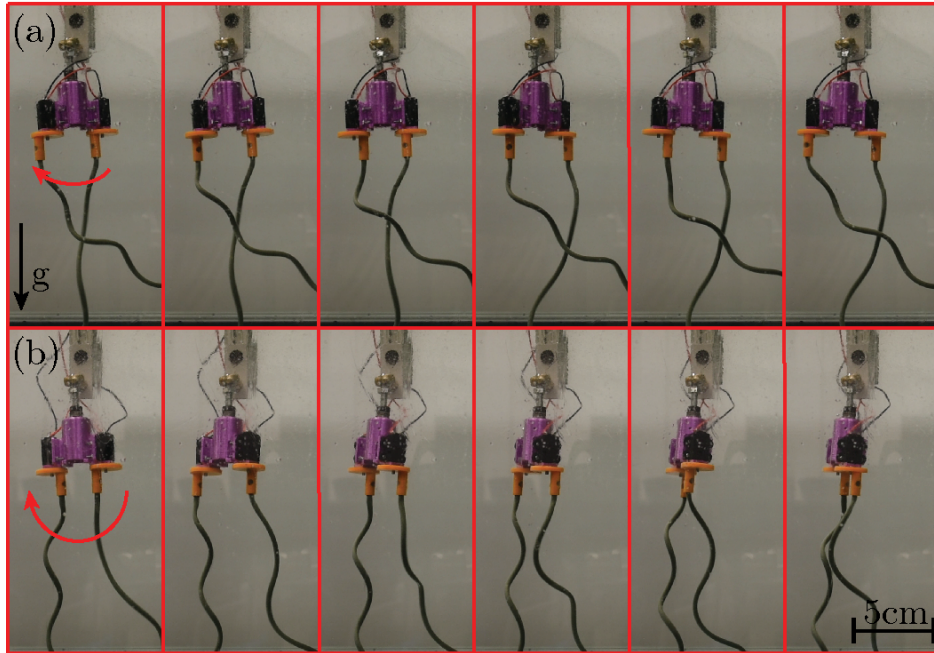


Figure 3.6: Attitude adjustment of bi-flagellated robot in the viscous fluid (glycerin). (a) Rotation in the horizontal plane (pointing opposite to gravity direction). (b) Rotation in the vertical plane (pointing out to the paper).

The paper is structured as follows. Section 3.3.2 provides a comprehensive overview of the experimental setup and robot structure, offering valuable insights into the practical aspects of our study. In Section 3.3.3, we delve into the details of the computational framework, which integrates various methodologies, including the Discrete Elastic Rod (DER) [8], Regularized Stokeslet Segments (RSS) [9], and Implicit Contact Model (IMC) [10]. This section also addresses the specific considerations for modeling the rotational dynamics of the bi-flagellated robot. Moving on to Section 3.3.4, we present a fundamental analysis of the attitude adjustment, encompassing critical aspects such as the reaction forces, the stability of Euler angles, and the range of attainable attitudes for the robot. This section delves into the core findings of our work, providing profound control application of the bi-flagellated robot. Lastly, in Section 3.3.5, we conclude our work by summarizing our key discoveries and outlining potential directions for future research in this exciting field.

3.3.2 Experiment Setup

3.3.2.1 Robotic structure

The robot illustrated in Figure 3.7(a) consists of an assembly base and two helical flagella connected to motor shafts. The base, with mass m of 94 g, features a central cylinder with a radius r_h of 13.5 mm and a height h of 37.5 mm. Two small brushed DC motors, rated at 6 V voltage and capable of a stall current of 1.5 A, are positioned on opposite sides of the cylinder at a distance d of 45 mm. The motors are powered through pulse-width modulation (PWM) signals from a microcontroller (Arduino Nano). Additionally, the robot is equipped with an inertial measurement unit (IMU) (MPU6050) located on the bottom plane of the base. This IMU tracks the robot's orientation in terms of Euler angles, including yaw ψ , pitch θ , and roll ϕ . These angles are defined in the body frame $x_B - y_B - z_B$ and are illustrated in Figure 3.7(b).

Referring to Figure 3.8, the soft flagella are fabricated using VPS material (Elite Double 32). To modulate the silicone elastomer's density, the base and catalyst are blended with iron powder. The resulting mixture is then injected into pre-shaped PVC tubes, and after

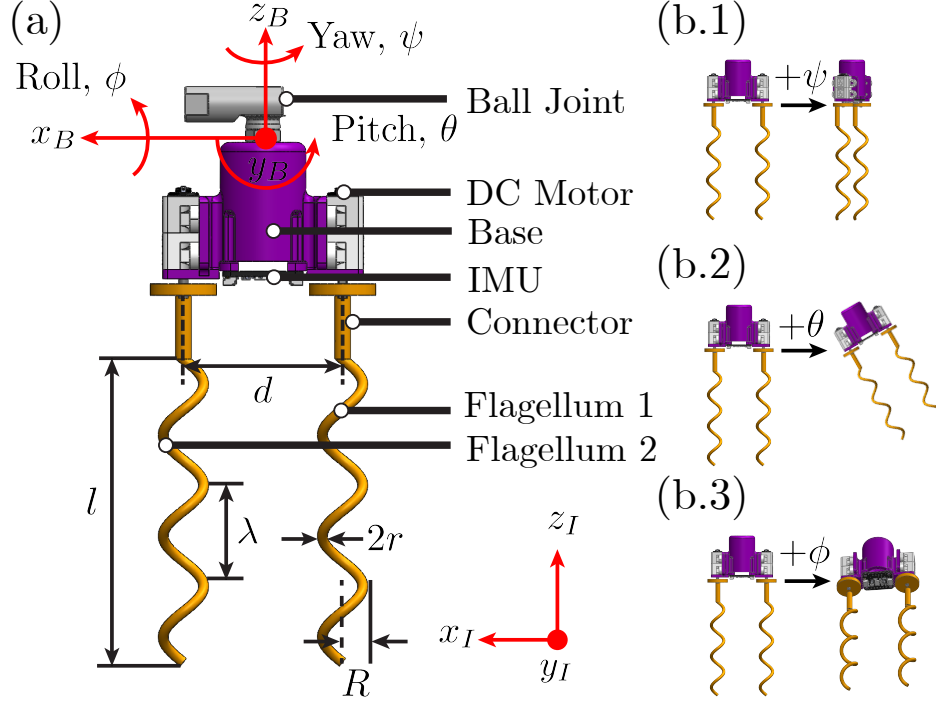


Figure 3.7: Robot schematic and attitude representation. (a) The bi-flagellated robot comprises an assembly base and two identical soft flagella. Each flagellum with length l , cross-sectional radius r , helix pitch λ , and helix radius R is actuated by a miniature brushed DC motor. The robot is affixed to a ball joint to reorient. (b) Yaw angle ψ (b.1), Pitch angle θ (b.2), and Roll angle ϕ (b.3) signifies the angular displacement around the vertical axis z_B , lateral axis y_B , and longitudinal axis x_B , respectively.

tube slitting, the soft flagellum is assembled onto the robot base. These helical flagella have specific dimensions, including a fixed cross-sectional radius r of 6 mm, a helix radius R of 8.89 mm, a helix pitch λ of 76 mm, and a helix axial length l of 196.2 mm. The silicone composite exhibits a slender profile, possessing a Young's modulus E of 1.255×10^6 Pa, Poisson's ration ν of 0.5, and density ρ of 1450 kg/m³. Each flagellum is actuated within a range of 0 to 90 rpm, ensuring that the Reynolds number Re remains below 0.1, as defined by the formula $Re = \rho\omega Rr_0/\mu$.

3.3.2.2 Experimental platform

The experimental setup, as depicted in Figure 4.1, includes a glycerin tank measuring 122 cm in length, 45 cm in width, and 51.5 cm in height. Glycerin, chosen for its density ρ_f of

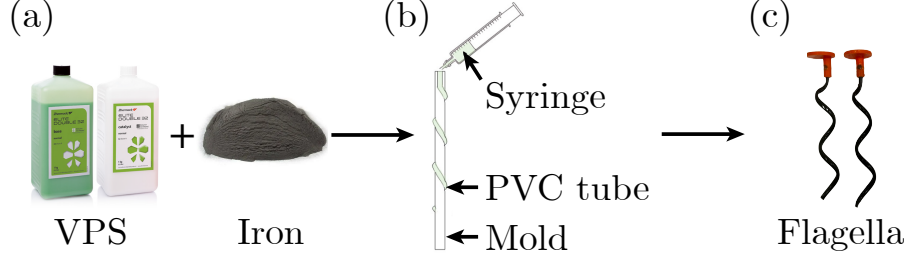


Figure 3.8: Fabrication of a soft flagellum. (a) Step 1: blending VPS with iron powder to precisely calibrate the density of the silicone composite. (b) Step 2: using a syringe to inject the mixture into the PVC tube preformed by a mold. (c) Step 3: slitting the tube to extract the silicone elastomer.

1.26 g/ml and viscosity μ of 1 Pa·s at 25° Celsius, serves as the surrounding viscous medium. To restrict the robot's motion, it is secured in place by a positioning frame featuring a ball joint that permits limited rotational movement. This setup ensures that the IMU located on the robot's central axis can accurately measure yaw, pitch, and roll angles over time. The IMU data is transmitted to an external microcontroller, responsible for both data processing and the independent control of the DC motors, enabling us to adjust their rotational speed and direction.

3.3.3 Numerical method

The computational framework for a bi-flagellated system comprises three key components: (1) the Discrete Elastic Rods method for modeling the nonlinear structural elasticity, resulting in flagella deformation and buckling, (2) the Regularized Stokeslet Segments method for handling long-range hydrodynamics, resulting in fluid interaction induced by flagella rotation, and (3) the Implicit Contact Model for realistic contact interactions, resulting in flagella bundling. To simulate the dynamic behavior of the system, we incorporate all the reaction forces and moments generated by these methods into the governing equation of rotation. This comprehensive approach enables an effective modeling and analysis of the system's motion response.

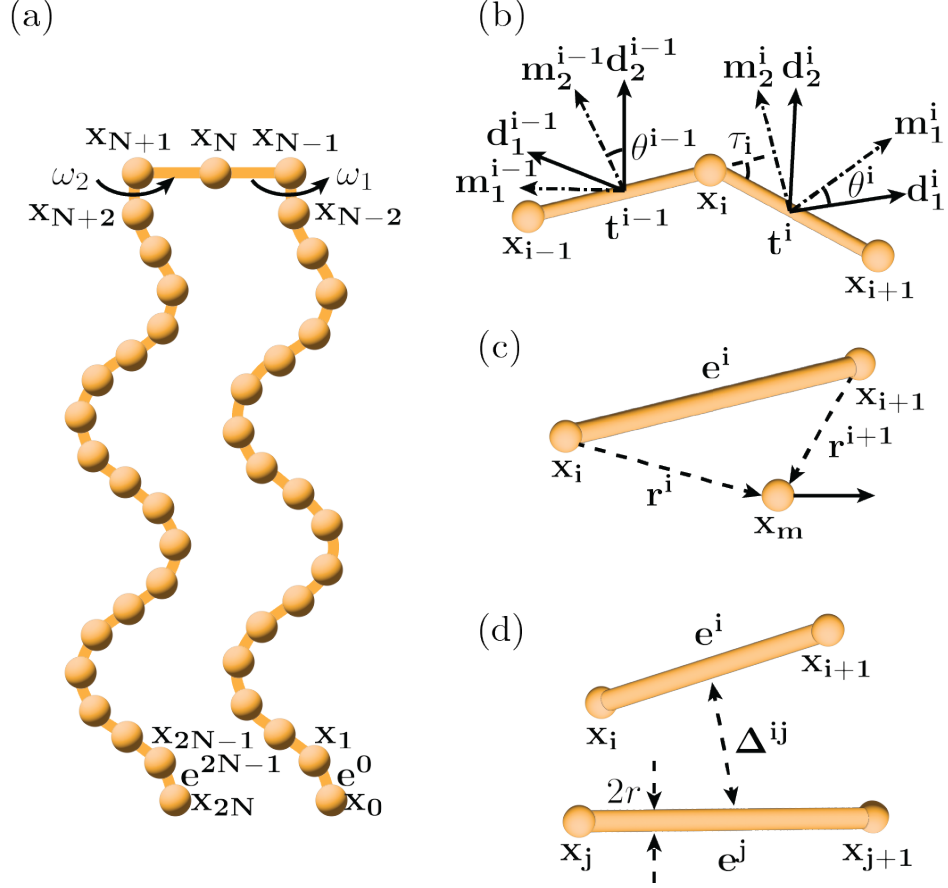


Figure 3.9: Computation framework of bi-flagellated robot. (a) Discretization scheme. Each soft flagellum is discretized into N nodes (denoted as $\mathbf{x}_i$) and $N - 1$ edges (denoted as $\mathbf{e}^i$), and connected to the base node $\mathbf{x}_N$. (b) Representation of material frame $([\mathbf{m}_1^{i-1}, \mathbf{m}_2^{i-1}, \mathbf{t}_{i-1}])$ and $([\mathbf{m}_1^i, \mathbf{m}_2^i, \mathbf{t}_i])$ and reference frame $([\mathbf{d}_1^{i-1}, \mathbf{d}_2^{i-1}, \mathbf{t}_{i-1}])$ and $([\mathbf{d}_1^i, \mathbf{d}_2^i, \mathbf{t}_i])$. The signed angle from $\mathbf{d}_1^i$ to $\mathbf{m}_1^i$ is θ^i and twist at node $\mathbf{x}_i$ is $\tau_i = \theta^i - \theta^{i-1}$. (c) Regularized Stokeslets Segments. This method establishes a connection between the velocity $\dot{\mathbf{x}}_m$ at a point $\mathbf{x}_m$ and the forces acting on the nodes $\mathbf{x}_i$ and $\mathbf{x}_{i+1}$. (d) Implicit contact model. The artificial contact energy is defined by the minimal distance Δ_{ij} between two edges $\mathbf{e}^i$ and $\mathbf{e}^j$.

3.3.3.1 Discrete elastic rods

Figure 3.9(a) provides a discrete schematic of the bi-flagellated robot. In our modeling approach, each flagellum is represented using Kirchhoff's elastic rod, commonly employed in computer graphics. The DER algorithm discretizes each Kirchhoff rod into N nodes with the segment length Δl of 5 mm, denoted as $\mathbf{x}_0, \mathbf{x}_1, \dots, \mathbf{x}_{N-1}$ and $\mathbf{x}_{N+1}, \mathbf{x}_1, \dots, \mathbf{x}_{2N}$. To simulate the rotation dynamics of the robot, we introduce an additional node, $\mathbf{x}_N$, which represents the robot base and connects with the endpoints of the two rods. Notably, there is

no relative motion among node $\mathbf{x}_{N-1}$, $\mathbf{x}_{N+1}$, and $\mathbf{x}_N$. Consequently, the system comprises a total of $2N + 1$ nodes (denoted as $\mathbf{n}_x$), characterized by $\mathbf{X} = \{\mathbf{x}_0, \mathbf{x}_1, \dots, \mathbf{x}_{2N}\}$, corresponding to a total of $2N$ edge vectors (denoted as $\mathbf{n}_e$), expressed as $\mathbf{E} = \{\mathbf{e}^0, \mathbf{e}^1, \dots, \mathbf{e}^{2N-1}\}$, with $\mathbf{e}^i = \mathbf{x}_{i+1} - \mathbf{x}_i$.

As depicted in Figure 3.9(b), each edge $\mathbf{e}^i$ is associated with two sets of orthonormal frames, responsible for accounting for rotation. These frames include a reference frame, denoted as $\{\mathbf{d}_1^i, \mathbf{d}_2^i, \mathbf{t}^i\}$, and a material frame, expressed as $\{\mathbf{m}_1^i, \mathbf{m}_2^i, \mathbf{t}^i\}$. Notably, both frames share the tangent vector $\mathbf{t}^i = \mathbf{e}^i/|\mathbf{e}^i|$ as one of their directors. The reference frame is initialized at time $t = 0$ and is updated at each time step through time-parallel transport. In contrast, the material frame can be calculated based on a scalar twist angle, θ^i .

Consequently, for each flagellum (in this case, we focus on flagellum 1 for simplicity), the spatial positions of nodes, $\mathbf{x}_i = (x_i, y_i, z_i)$, and the twist angle θ^i collectively constitute the degrees of freedom (DOF) vector, which has a size of $4N - 1$, i.e.,

$$\mathbf{q} = [\mathbf{x}_0, \theta^0, \mathbf{x}_1, \dots, \mathbf{x}_{N-2}, \theta^{N-2}, \mathbf{x}_{N-1}]^T,$$

where superscript T represent transpose. The equations of motion (EOM) of the i -th DOF is

$$m_i \ddot{q}_i + \frac{\partial E_{\text{potential}}}{\partial q_i} + f^{\text{external}} = 0,$$

where $E_{\text{potential}}$ is formulated in the remainder of this part, f^{external} consists of external hydrodynamics forces $\mathbf{f}^h$ and contact force $\mathbf{f}^c$ that are addressed in the Section 3.3.3.2 and 3.3.3.3. Note that m_i stands for mass (unit: kg) if q_i corresponds to spatial position and moment of inertia (unit: kg-m²) if q_i corresponds to twist angle.

According to Kirchhoff's rod theory, the potential energy can be assumed to be a sum of elastic stretching, bending, and twisting energies, i.e.,

$$\begin{aligned} E_{\text{potential}} &= E^s + E^b + E^t \\ &= \sum_{i=0}^{N-2} E_{\mathbf{k}}^s + \sum_{i=1}^{N-2} E_{\mathbf{k}}^b + \sum_{i=1}^{N-2} E_{\mathbf{k}}^t. \end{aligned}$$

Stretching energy. The axial stretch, ϵ , of edge $\mathbf{e}^{\mathbf{i}}$ is

$$\epsilon^{\mathbf{i}} = \frac{\|\mathbf{e}^{\mathbf{i}}\|}{\|\bar{\mathbf{e}}^{\mathbf{i}}\|} - 1,$$

where $\|\bar{\mathbf{e}}^{\mathbf{i}}\|$ is length of the edge in undeformed state. The stretching energy along edge $\mathbf{e}^{\mathbf{k}}$ is

$$E_{\mathbf{i}}^s = \frac{1}{2}EA(\epsilon^{\mathbf{i}})^2\|\bar{\mathbf{e}}^{\mathbf{i}}\|,$$

where $EA = E\pi r^2$ is the stretching stiffness, E is the Young's modulus, and r is the rod radius.

Bending energy. Bending strain is measured at each node $\mathbf{x}_{\mathbf{i}}$ through the curvature binormal vector

$$(\kappa \mathbf{b})_{\mathbf{i}} = \frac{2\mathbf{e}^{\mathbf{i}-1} \times \mathbf{e}^{\mathbf{i}}}{\|\mathbf{e}^{\mathbf{k}-1}\|\|\mathbf{e}^{\mathbf{i}}\| + \mathbf{e}^{\mathbf{i}-1} \cdot \mathbf{e}^{\mathbf{i}}}.$$

The scalar curvatures along the first and second material directors using the curvature binomial are

$$\begin{aligned}\kappa_{\mathbf{i}}^{(1)} &= \frac{1}{2}(\mathbf{m}_2^{\mathbf{i}-1} + \mathbf{m}_2^{\mathbf{i}}) \cdot (\kappa \mathbf{b})_{\mathbf{i}}, \\ \kappa_{\mathbf{i}}^{(2)} &= \frac{1}{2}(\mathbf{m}_1^{\mathbf{i}-1} + \mathbf{m}_1^{\mathbf{i}}) \cdot (\kappa \mathbf{b})_{\mathbf{i}}.\end{aligned}$$

The bending energy at node $\mathbf{x}_{\mathbf{k}}$ is

$$E_{\mathbf{i}}^b = \frac{1}{2} \frac{EI}{\Delta l_{\mathbf{i}}} [(\kappa_{\mathbf{i}}^{(1)} - \bar{\kappa}_{\mathbf{i}}^{(1)})^2 + (\kappa_{\mathbf{i}}^{(2)} - \bar{\kappa}_{\mathbf{i}}^{(2)})^2],$$

where $\Delta l_{\mathbf{i}} = \frac{1}{2}(\|\bar{\mathbf{e}}^{\mathbf{i}-1}\| + \|\bar{\mathbf{e}}^{\mathbf{i}}\|)$ is the Voronoi length for the $\mathbf{i}$ -th node, $\bar{\kappa}_{\mathbf{i}}^{(1)}$ and $\bar{\kappa}_{\mathbf{i}}^{(2)}$ are the material curvatures in undeformed state, and $EI = E\pi r^4/4$ is the bending stiffness.

Twisting energy. The twist between two consecutive edges at node $\mathbf{x}_{\mathbf{i}}$ is

$$\tau_{\mathbf{i}} = \theta^{\mathbf{i}} - \theta^{\mathbf{i}-1} + \Delta m_{\mathbf{i},\text{ref}},$$

where $\Delta m_{\mathbf{i},\text{ref}}$ is the reference twist, which is the twist of the reference frame as it moves

from the $(\mathbf{i} - \mathbf{1})$ -th edge to the $\mathbf{i}$ -th edge. The twisting energy is

$$E_{\mathbf{i}}^t = \frac{1}{2} \frac{GJ}{\Delta l_{\mathbf{i}}} (\tau_{\mathbf{i}} - \bar{\tau}_{\mathbf{i}})^2,$$

where $\bar{\tau}_{\mathbf{i}}$ is the twist along the centerline in the undeformed state, $GJ = G\pi r^2/2$ is the twisting stiffness, and G is the shear modulus.

3.3.3.2 Regularized Stokeslets Segments

We employ the Regularized Stokeslets Segments method to model the long-range hydrodynamics interaction between two flagella and viscous fluids. This method establishes a linear relationship between the velocity vector $\mathbf{U}$ (size $3\mathbf{n}_{\mathbf{x}}$) at the node set $\mathbf{X}$ and the hydrodynamic force vector $\mathbf{F}$ (size $3\mathbf{n}_{\mathbf{x}}$) acting upon them. This relationship is defined by a geometry-associated matrix $\mathbf{A}$ (size $3\mathbf{n}_{\mathbf{x}} \times 3\mathbf{n}_{\mathbf{x}}$), i.e.,

$$\mathbf{U} = \mathbf{A}\mathbf{F}. \quad (3.1)$$

Referring to Figure 3.9(c), the RSS method characterizes the flow arising from a singular point force. It establishes a connection between the velocity $\dot{\mathbf{x}}_{\mathbf{m}}$ at a point $\mathbf{x}_{\mathbf{m}}$ and the forces exerted by each node on the fluid, i.e.,

$$8\pi\mu\dot{\mathbf{x}}_{\mathbf{m}} = \sum_{\mathbf{i}=0}^{\mathbf{N}-2} (\mathbf{A}_1^{\mathbf{i}}\mathbf{f}_{\mathbf{i}}^h + \mathbf{A}_2^{\mathbf{i}}\mathbf{f}_{\mathbf{i}+1}^h),$$

where $\mathbf{f}_{\mathbf{i}}^h$ (size 3) signifies the force vector corresponding to the force applied by the $\mathbf{i}$ -th node onto the fluid. This force is equal in magnitude but opposite in direction to the hydrodynamic force acting on the $\mathbf{i}$ -th node. The construction of matrices $\mathbf{A}_1^{\mathbf{i}}$ and $\mathbf{A}_2^{\mathbf{i}}$ follows the procedures in [59].

Matrix $\mathbf{A}$ is then assembled from matrices $\mathbf{A}_1^{\mathbf{i}}$ and $\mathbf{A}_2^{\mathbf{i}}$ with proper rearrangement. At each time step, by calculating the positions $\mathbf{X}$ and velocities $\mathbf{U}$ using the DER method, we initially derive matrix $\mathbf{A}$. Subsequently, we employ the Equation 3.1 to evaluate the

hydrodynamic forces at each node and compute the resultant forces of each flagellum, i.e.,

$$\mathbf{F}^1 = - \sum_{i=0}^{N-1} \mathbf{f}_i^h, \mathbf{F}^2 = - \sum_{i=N+1}^{2N} \mathbf{f}_i^h.$$

Lastly, we apply $\mathbf{F}^1 = [F_x^1, F_y^1, F_z^1]$ and $\mathbf{F}^2 = [F_x^2, F_y^2, F_z^2]$ on node $\mathbf{x}_{N-1}$ and $\mathbf{x}_{N+1}$ to study the rotation dynamics in Section 3.3.3.4.

3.3.3.3 Implicit contact model

In light of the physical rod-on-rod interaction between two flagella, we introduce the Implicit Contact Model to handle the contact interactions between the impenetrable edges. IMC is a fully implicit penalty-based frictional contact method that has demonstrated its capability to accurately capture challenging contact scenarios, including instances like flagella bundling. In contrast to previous studies [27], which employed an adaptive contact model (Euler forward) in conjunction with DER, we have opted to replace this explicit approach with IMC for two compelling reasons: (i) The implicit (Euler backward) implementation of IMC can effectively circumvent numerical convergence issues, enhancing the stability and reliability of the simulation. (ii) The model's capacity to accommodate larger time steps without compromising accuracy greatly accelerates the simulation, contributing to computational efficiency and a more streamlined modeling process.

Referring to Figure 3.9(d), we examine an edge-to-edge contact pair denoted as $\mathbf{x}_{ij} := (\mathbf{x}_i, \mathbf{x}_{i+1}, \mathbf{x}_j, \mathbf{x}_{j+1})$ from the node set $\mathbf{X}$. Detailed in [10], the formulation of frictional contact forces follows the subsequent steps: (i) computing the minimal distance Δ^{ij} between two edges $\mathbf{e}^i$ and $\mathbf{e}^j$, (ii) assessing the contact energy $E(\Delta^{ij}, \delta)$ with respect to the rod cross-sectional radius r and contact distance tolerance δ of 0.01 mm, and (iii) determining the contact forces as the negative gradient of the contact energy, i.e.,

$$\mathbf{f}_{ij}^c = -k \nabla_{\mathbf{x}} E(\Delta^{ij}, \delta),$$

where k is the contact stiffness to ensure the non-penetration.

3.3.3.4 Rotation dynamics

Building upon the previously outlined computational framework, we delve into the development of the rotational dynamics for the bi-flagellated robot, which is attached to a ball joint. The primary objective of this analysis is to explore the potential for adjusting the robot's attitude. To mitigate the common singularity issues associated with Euler angles, we opt for quaternions represented as $\mathbf{q}_o = [q_w, q_x, q_y, q_z] \in \mathbb{R}^4$ and angular velocity components along the body axes x_B, y_B, z_B , denoted as $\boldsymbol{\omega}_o = [\omega_x, \omega_y, \omega_z] \in \mathbb{R}^3$, to describe the robot's orientation. If needed, the Euler angles $[\psi, \theta, \phi]$ can be easily converted by quaternions, e.g., `quat2eul` function available in MATLAB. The governing equation for rotation is articulated as follows

$$\begin{aligned} \frac{d\mathbf{q}_o}{dt} &= \frac{1}{2} \mathbf{q}_o \otimes \begin{bmatrix} 0 \\ \boldsymbol{\omega}_o \end{bmatrix}, \\ \frac{d\boldsymbol{\omega}_o}{dt} &= \mathbf{J}^{-1}(\mathbf{M} - \boldsymbol{\omega}_o \times (\mathbf{J}\boldsymbol{\omega}_o)), \end{aligned} \quad (3.2)$$

where $\mathbf{J}$ represents the moment of inertia matrix, $\otimes$ signifies quaternion multiplication, and $\mathbf{M} = [M_x, M_y, M_z]$ corresponds to the external torque applied to the robot's base along the x_B, y_B, z_B axes.

Taking into account the impact of hydrodynamics and inertia, we derive the expression for the external torque as

$$\begin{aligned} M_x &= (F_x^1 - F_x^2) \times 0 + M_{hx}, \\ M_y &= (F_z^1 - F_z^2) \times \frac{d}{2} + M_r + M_{hy}, \\ M_z &= (F_y^1 - F_y^2) \times \frac{d}{2} + M_{hz}, \end{aligned}$$

where $F_x^1, F_x^2, F_z^1, F_z^2, F_y^1, F_y^2$ are the flagella forces as described in Section 3.3.3.2. It's worth noting that although F_x^1, F_x^2 have nontrivial values, they do not effectively apply a torque to the robot since the lever arm is zero. Furthermore, M_r represents the righting moment, which seeks to restore a tilted robot to an upright position due to the mass center shift r_m

measuring 2 mm, with expressioin

$$M_r = mgr_m \sin \theta, \quad (3.3)$$

where m is the base mass, θ is the titled pitch angle, and g is gravitational acceleration.

Lastly, we consider the viscous drag exerted on the robot base due to the angular velocity ω_o , i.e.,

$$\begin{aligned} M_{hx} &= -2\pi\mu C_{rx} r_h^3 \omega_x, \\ M_{hy} &= -2\pi\mu C_{ry} r_h^3 \omega_y, \\ M_{hz} &= -2\pi\mu C_{rz} r_h^3 \omega_z, \end{aligned}$$

where C_{rx}, C_{ry}, C_{rz} accounts for the nonsphericity of the robot's base and can be empirically determined, μ is the viscosity, and r_h is the base radius.

3.3.4 Results and discussion

3.3.4.1 Analysis of force reaction

The interaction between the flagella and the surrounding fluid medium leads to a distinct force reaction, illustrated in Figure 3.10(a). The flagella exhibit not only a stable propulsion thrust F_z (blue curves), but also induce periodic forces F_x and F_y (orange and yellow curves, respectively).

The propulsion thrust F_z^1 and F_z^2 are generated by the two flagella at a distance d . This thrust gives rise to a torque that effectively changes the pitch angle θ within the $x_B - z_B$ plane. Simultaneously, the periodic forces, F_y^1 and F_y^2 , possessing different phases and magnitudes, exert a torque responsible for altering the yaw angle ψ within the $x_B - y_B$ plane.

However, it is important to note that due to the points of reaction being situated along the x_B axis, the forces F_x^1 and F_x^2 do not contribute to changes in the roll angle ψ within the $y_B - z_B$ plane. As a result, we deduce that in the context of bi-flagellated robots, the effective adjustment of attitude angles is limited to yaw and pitch. To modify the roll angle, additional flagella placed off the longitudinal axis x_B or alterations in the positioning angle

between the two flagella would be required.

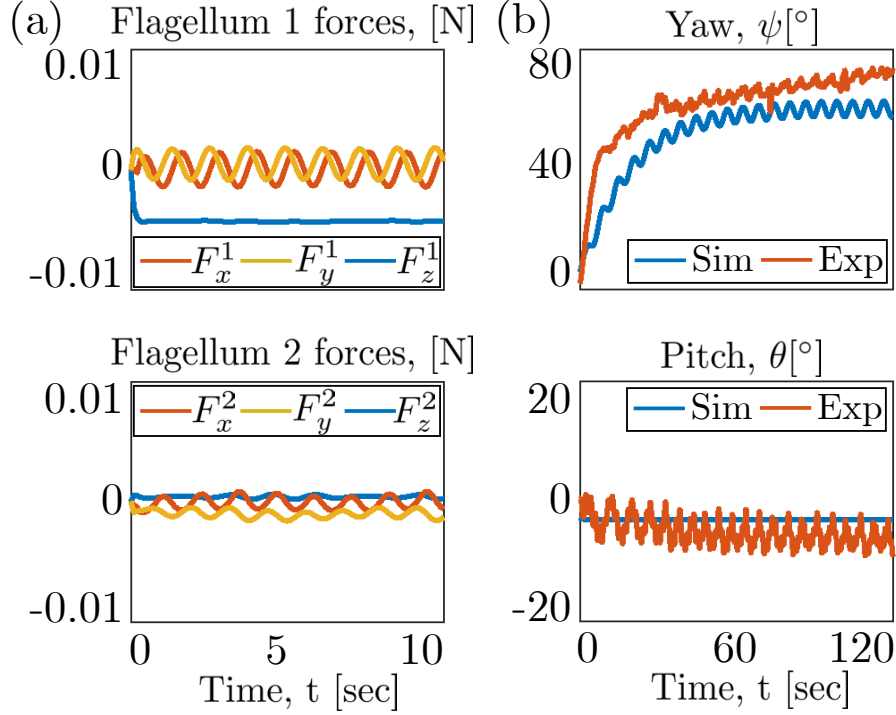


Figure 3.10: Time evolution of reaction forces and attitude angles under dual actuation of $\omega_1 = 50$ rpm and $\omega_2 = 10$ rpm. (a) Examination of Flagellum 1 (Top) and Flagellum 2 (Bottom) reveals that the vertical force component, F_z , rapidly stabilizes after an initial transition, while the horizontal forces, F_x and F_y , exhibit sinusoidal behavior. For a right-handed helix, the counter-clockwise rotation ($\omega > 0$) generates an downward propulsion force. (b) Yaw angle ψ and pitch angle θ eventually stabilize, reaching steady-state values of $\psi_{ss} = 60.76^\circ$ and $\theta_{ss} = -3.29^\circ$ in simulation, and $\psi_{ss} = 67.95^\circ$ and $\theta_{ss} = -4.96^\circ$ in experiment.

3.3.4.2 Stabilization of attitude angle

Using Equation 3.2, we investigate the time-dependent evolution of yaw and pitch angle while actuating two flagella with different speeds. Figure 3.10(b) displays the response of Euler angles when the force input follows the pattern presented in Figure 3.10(a). Notably, the two angles exhibit a convergence to stable values of 60.76° and -3.29° , respectively, which closely align with experimental results. In this section, we delve into an analysis of the factors contributing to the convergence observed in two attitude angles.

For the pitch angle θ , the impact of the righting moment M_r in Equation 3.3 becomes

particularly evident. The propulsion thrust does not lead to endless changes in the pitch angle θ , and instead, it contributes to the establishment of a stable equilibrium value associated with the system's intrinsic properties. Similarly, the evolution of the yaw angle ψ experiences constraints due to the presence of viscous drag in the fluid. This drag force acts as a restraint, preventing the yaw angle from continuously increasing and thereby contributing to the overall stability of the system.

As a consequence of these underlying principles, we observe that, for a given actuation of the two flagella, the robot's attitude always undergoes a gradual transition to a steady state, denoted as ψ_{ss} and θ_{ss} . This intriguing evolution provide valuable insights into the dynamic behavior of the bi-flagellated robot under the specified actuation conditions.

3.3.4.3 Visualization of attainable attitude

With the established convergence of attitude angles, we gain confidence in the robot's ability to attain a stable attitude over time. This stable attitude is defined in terms of the actuation speeds, ω_1 and ω_2 , providing insight into the attitude adjustment possibilities of the bi-flagellated robot.

In our simulation, we systematically vary ω_1 from -90 rpm to 90 rpm and ω_2 from 0 to 90 rpm in 10 rpm intervals, resulting in a total of 190 trials. In the experimental setup, we randomly select speed pairs (ω_1, ω_2) from the same range, totaling 75 trials. Each trial is conducted over an extended period, approximately 120 seconds, to ensure the emergence of attitude stability. To visualize the results, we present the steady-state values $|\psi_{ss}|$ and $|\theta_{ss}|$ (where $|\cdot|$ denotes the absolute value) in a colormap, using actuation speeds ω_1 and ω_2 as the x and y axes, respectively.

The results, depicted in Figure 3.11, unveil the robot's remarkable capability to adjust its yaw and pitch angles through specific actuation configurations. For instance, to achieve the maximum yaw angle, one should explore the upper right corner of Figure 3.11(a) with $|\psi_{ss}| = 180^\circ$ and $(\omega_1, \omega_2) = (+90, +90)$ rpm. Conversely, for maximum pitch angle, one should examine the upper left corner of Figure 3.11(b) with $|\theta_{ss}| = 16^\circ$ and $(\omega_1, \omega_2) = (-90, +90)$

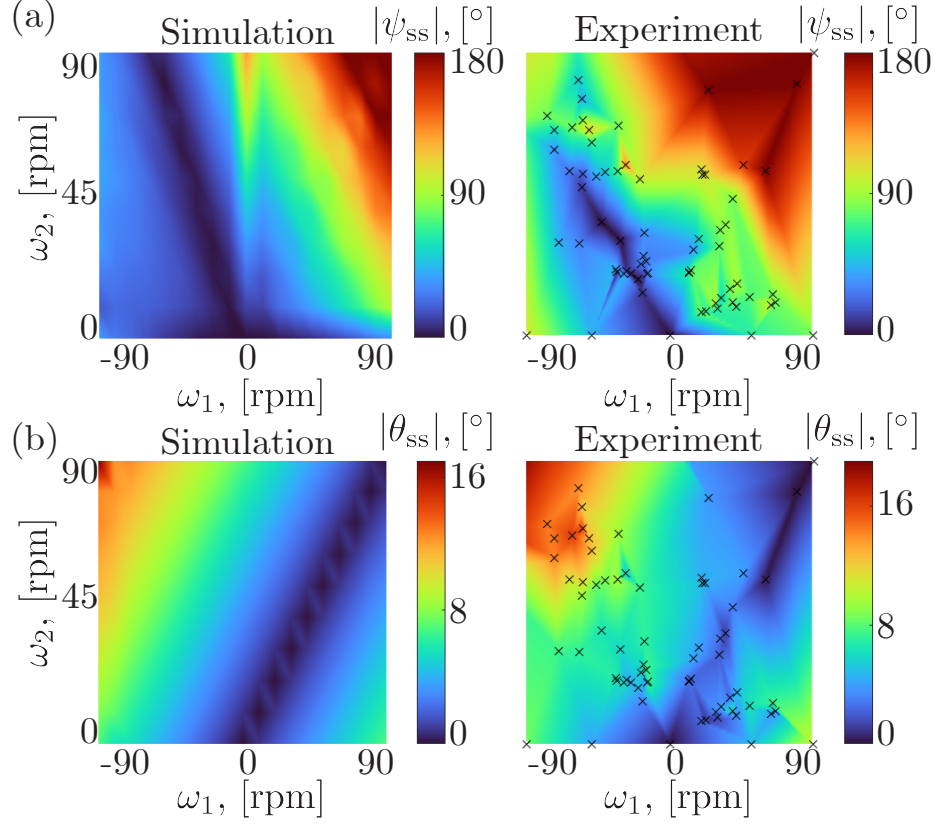


Figure 3.11: The steady-state values of (a) yaw angle $|\psi_{ss}|$ and (b) pitch angle $|\theta_{ss}|$ are depicted in response to actuation speeds ω_1 and ω_2 . The generated maps represent results from simulations (left) and experiments (right), based on 180 and 75 trials, respectively. $\times$ marks the experimental data points. Motor speed was controlled by varying PWM values and evaluated through videos.

rpm. This information allows us to predict the expected attitude for any given actuation. For example, when $(\omega_1, \omega_2) = (0, +45)$ rpm, the attitude will stabilize at $|\psi_{ss}| = 56.22^\circ$ and $|\theta_{ss}| = 3.45^\circ$.

Notably, two distinctive lines become evident in both experimental and simulation results: $\omega_2 = -\omega_1$ in Figure 3.11(a) and $\omega_2 = \omega_1$ in Figure 3.11(b). Along these lines, the steady state value $|\psi_{ss}|$ or $|\theta_{ss}|$ reaches a local minimum in their respective neighborhoods. This intriguing behavior can be interpreted as follows. The minimal yaw angle region ($\omega_2 = -\omega_1$) arises from the opposing forces F_y^1 and F_y^2 generated by the two flagella, which effectively cancel each other out and prevent any significant rotation torque from being applied to the robot. Conversely, for minimal pitch angle region ($\omega_2 = \omega_1$), the thrust forces F_z^1 and F_z^2

remain balanced, resulting in no flip torque being applied to the robot. Consequently, the pitch angle remains largely unaffected. Therefore, to achieve simultaneous changes in both yaw and pitch angles, an unbalanced speed for the two flagella is necessary.

In conclusion, the generated map not only illustrates all potential attitudes for the bi-flagellated robot but also serves as a valuable tool for pinpointing actuation speed. This map’s applicability extends to various aspects of robot design, including flagellum geometry, elasticity, fluid viscosity, and robot structure. For clarity, we highlight four dimensionless variables influencing the map’s pattern, including

$$\frac{\omega\mu l^4}{EI}, \frac{d}{l}, \frac{\lambda}{L}, \frac{R}{L}.$$

These insights provide a solid foundation for the efficient control of attitude through strategic actuation configurations, with wide applications including untethered robots requiring precise reorientation.

3.3.5 Conclusions and future work

In summary, this study delves into the attitude adjustment capabilities of a scaled up bi-flagellated system. The investigation begins with an analysis of force reactions resulting from the flagella’s interaction with the fluid, encompassing stable propulsion thrust and periodic forces. The research further explores the stabilization of the attitude angles, shedding light on the factors contributing to their convergence. A significant outcome is the visual representation of attainable attitudes in the generated map. Actuation speeds are systematically varied through simulations and experiments, illustrating the robot’s capacity to fine-tune its yaw and pitch angles in response to specific actuation configurations.

The study emphasizes the practical applicability of this map for designing and controlling bi-flagellated robots, offering valuable insights into critical design parameters. Ultimately, we conclude these findings lay a strong foundation for precise attitude control of untethered robots.

CHAPTER 4

Ring Structures for Peristaltic Transport

4.1 Overview

Peristalsis, a biologically inspired mechanism, plays a crucial role in locomotion and material transport in living systems. While extensively studied in nature, its application in soft robotics for handling and transporting objects has seen progress but remains limited. This study presents a pneumatic modular actuator, fabricated from silicone polymer, that is scalable, adaptable, and repairable in situ. The system integrates donut-shaped actuation modules capable of radial and axial inflation, coupled with real-time pressure feedback for synchronized control across multiple stacked modules. Experimental validation demonstrates the actuator’s ability to grasp and transport objects with diameters as small as 0.4 times its inner diameter at a speed of $2.08 \pm 0.07\text{mm/s}$. The system successfully handles a range of object materials, including deformable soft tubes, solid handheld levels, and irregularly shaped bundles of pens. This work advances peristaltic actuation on object transportation, enabling safe and reliable manipulation of deformable and irregularly shaped materials across various applications, such as underwater specimen delivery and field robotics operations.

4.2 Introduction

Peristalsis, defined as the involuntary, wave-like contraction and relaxation of circular and longitudinal muscles [33], is a widespread biological mechanism essential for various functions in animals and humans. It primarily enables locomotion and facilitates the transport of substances such as food, fluids, and waste. On one hand, soft-bodied animals, such as earth-

worms, sea cucumbers, snails, and certain types of fish, use peristalsis for locomotion [37]. They rely on wave-like muscular elongation and contraction along their bodies to propel themselves through their environment [38], such as soil, water, or along surfaces. On the other hand, peristalsis is fundamental for biological organisms to transport solid or liquid substances [36], facilitated by the propagation of rhythmic contraction waves along tubular structures [34], such as the esophagus, stomach, intestines, and ureter [39]. The process provides slow but stable [60] and adaptable transportation [61], minimizing energy consumption [47] and enabling movement through small or irregular openings [52] while maintaining functionality under varying conditions [35].

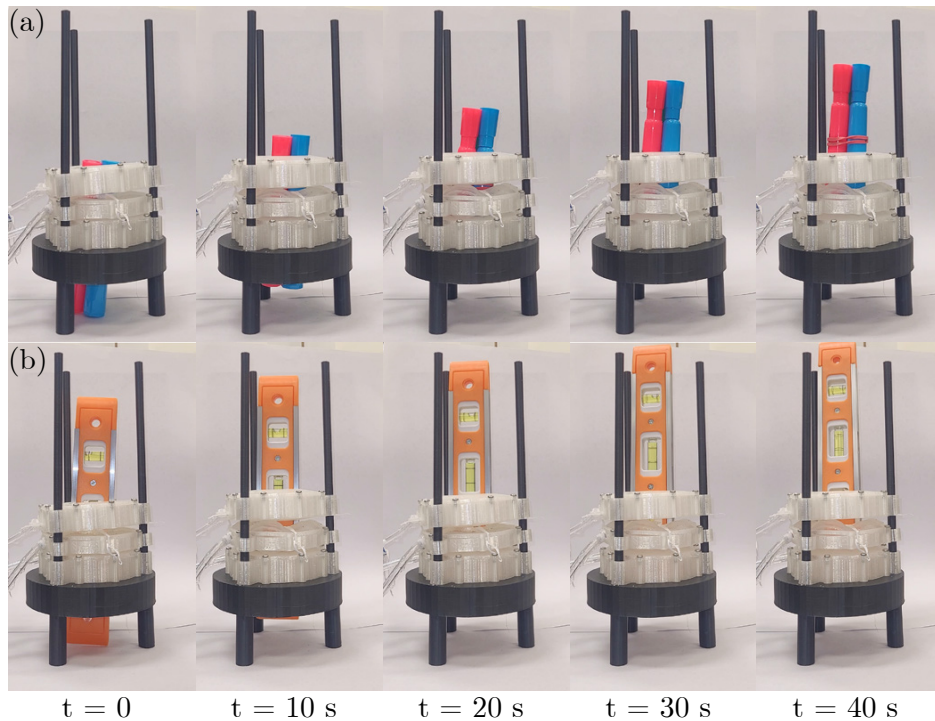


Figure 4.1: Demonstration of the actuator’s capability to transport irregular objects using a fundamental unit with three stacked modules: (a) a bundle of marker pens and (b) a handheld leveling tool.

With recent advancements in soft robotics, researchers have increasingly focused on developing soft robots and bio-inspired systems that replicate biological mechanisms and behaviors [1], [2]. To leverage the benefits of a soft, compliant body and the peristalsis mechanism, worm-like crawling robots have been developed using various actuation, fabrication, and control methods [53]. These studies highlight the advantages of the peristaltic

mechanism, including its adaptability to diverse environments, high compliance, and fault tolerance, ensuring reliability in complex systems. However, most existing works focus on locomotion rather than object transport applications. In particular, studies on peristaltic transport system often center on monolithic devices for biological applications. For example, Dirven et al. [45] developed a pneumatically actuated peristaltic device made from silicone rubber, which mimics esophageal swallowing and enables continuous, compliant transport through occlusive pressure waves. Likewise, Pend et al. [46], [48] created a pneumatic, large intestine-inspired soft robot that uses McKibben muscles for axial and circumferential contractions, facilitating object transport via traveling waves. In a similar vein, Uzawa et al. [49] and Kawano et al. [50] proposed the peristaltic transport system for space toilets that uses negative pressure to move biological waste. However, these prior works are constrained by manufacturing complexity, high costs, and limited adaptability to objects of varying sizes, such as compact and elongated shapes. In contrast, modular systems offer clear benefits, including enhanced portability, rapid deployment, and flexibility to adapt to various tasks, making them more sustainable in a wide range of applications, such as space station operations and search and rescue missions. Additionally, unlike monolithic designs where the failure of any component leads to system-wide breakdown, modular systems allow for the replacement or repair of only the affected module. This reduces downtime, resource consumption, and enhances operational continuity.

Inspired by studies on toroidal actuators, we recognize the potential for developing a modular system that leverages both radial and axial expansion – not only for gripping applications through radial contraction but also for propagation via axial expansion. Dang et al. [62] introduced a ring-shaped pneumatic actuator capable of radial contraction, demonstrating its effectiveness in gripping objects such as eggs and beakers. Hashem et al. [51] further explored this concept, showcasing its ability to grip delicate items – such as bellows, cherry tomatoes, and soft candy – by adjusting the pneumatic system’s loading conditions. Root et al. [54] utilized a hydrostatic mechanism to achieve controlled gripping, catching, and object conveyance. Zang et al. [55] presented a torus-like self-adaptive grasper capable of handling objects of unknown shapes, sizes, and softness. By adjusting the rolling direction

and length of the torus skin, the system could engulf, swallow, and release objects. Perez et al. [7] developed a toroidal actuator with both axial and radial expansion; however, its radial expansion is outward rather than inward, limiting its ability to achieve internal object transport. Despite these advancements, there remains a gap in the literature regarding the use of successive radial and axial expansions to enable controlled peristaltic motion for object transport.

To bridge this gap, we propose a system that leverages sequential radial and axial deformations to enable continuous and adaptive object transport, as Figure 4.1 shown. Unlike conventional soft grippers that rely on discrete pick-and-place operations, our peristaltic transport system facilitates seamless handling without the need for precise positioning, making it well-suited for manipulating objects of varying shapes and sizes. The system is designed to be modular, allowing for adjustable lengths tailored to specific tasks. Its modularity ensures easy transport and assembly for rapid deployment while maintaining resilience—ensuring that the failure of a single module does not compromise the entire operation. Our contributions include:

1. The design of a donut-shaped actuation module, along with the development of a modular actuator system.
2. A real-time pressure feedback control strategy to ensure smooth object handling and transportation.

Additionally, we provide a tutorial video¹ to facilitate reproduction and publicly release all simulation data² for future research and applications.

The paper is structured as follows: Section 4.3 provides an overview of the actuator design and fabrication. Section 4.4 delves into the peristaltic transportation mechanism and its control strategies. Section 4.5 presents the experimental and simulation results on the peristaltic actuator. Finally, Section 4.6 provides concluding remarks and future directions.

¹<https://youtu.be/uWucfxMYyb0>

²https://github.com/StructuresComp/Peristaltic_Actuator.git

4.3 Peristaltic Actuator

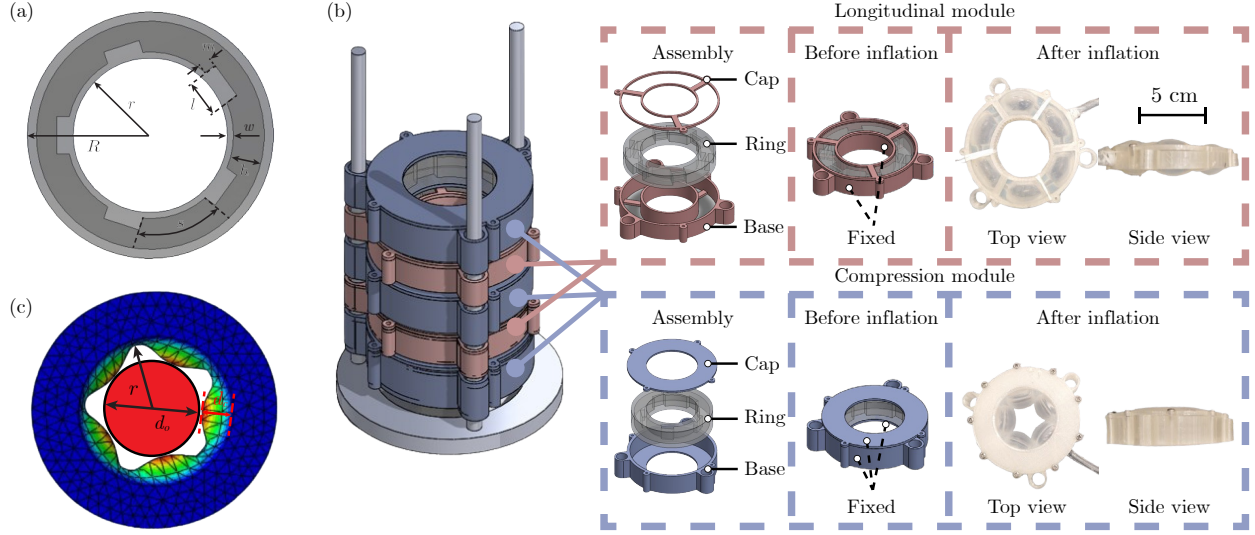


Figure 4.2: Design of the actuation module. (a) Each module is donut-shaped with outer radius (R), inner radius (r), and multiple air chambers defined by length (s), width (h), spacing (l), and wall thickness (t). Chambers are interconnected by air tunnels with step height (m). (b) The actuator station consists of stacked modules. In the longitudinal module, the casing permits inflation along the top and bottom surfaces, while in the compression module, inflation is restricted to the inner surface. (c) Finite element mesh of the actuation module, showing the maximum radial inflation displacement (d_c). The object is characterized by its maximum cross-sectional dimension d_o .

This section details the design and fabrication of the presented bio-inspired peristaltic actuator. The actuator features two distinct types of inflation modules: a compression module, which produces radial displacement, and a longitudinal module, which generates axial displacement. The modular design enables easy stacking of multiple units in any configuration or sequence, allowing for enhanced scalability and adaptability.

4.3.1 Design and Material Specifications

Each actuation module is composed of donut-shaped soft ring containing multiple internal air chambers, as shown in Figure 4.2(a), encased in a 3D-printed outer casing. To simplify fabrication, both types of actuation modules – compression and longitudinal – utilize a similar ring design as shown in Figure 4.2(b). The strain-limiting casing restricts the deformation of the soft ring in specific directions: the compression module is prevented from axial deforma-

tion, while the longitudinal module is constrained from radial deformation. As a result, the compression module only inflates radially to grasp an object, while the longitudinal module inflates axially to transport it along the axis of the stacked modules, hereafter referred to as the “station.” This modular approach allows for low-cost replacements in the event of partial module failures, eliminating the need for multiple specialized molds.

The inclusion of multiple air chambers is essential for promoting uniform radial inflation. Using a single air chamber may cause visible inhomogeneous deformation, resulting in a loss of axisymmetry due to the self-contact. By incorporating multiple chambers, we also minimize the risk of defects in one chamber affecting the inflation of others.

The rings are fabricated from silicone rubber (Smooth-On Ecoflex™ 00-45 Near Clear™), selected for its low cost and hyperelastic behavior with relatively low stiffness (Shore 00 hardness 45) that facilitates easy inflation. Its hyperelastic nature permits deformations over numerous cycles without permanent shape change, enabling effective pneumatic actuation and ensuring consistent performance without degradation over time. The outer casing is constructed from Thermoplastic Polyurethane (TPU) (Giantarm™ 95A TPU Filament, 1.75mm, Clear), which has a higher stiffness (Shore A hardness 95). TPU serves as a strain-limiting layer, providing the necessary structural support to withstand the weight of stacked modules in the station. Without TPU, the stacked soft rings would be unable to maintain the station’s overall shape due to their own weight.

Table 4.1: Geometrical parameters

Description	Symbol	Value	Unit
Outer radius	R	40	mm
Inner radius	r	25	mm
Step height	m	4	mm
Chamber spacing	l	12	mm
Wall thickness	w	2	mm
Chamber length	s	28.8	mm
Chamber width	h	10	mm
Object dimension	d_o	20 – 50	mm

4.3.2 Fabrication and Assembly

The base and cap of the TPU casings, are printed using a 3D printer (Original Prusa™ i3 MK3S+ 3D Printer). Following this, we fabricate the soft rings through a step-by-step casting and curing process, as illustrated in [Figure 4.3](#). To prepare the mixture, combine Part A and Part B of Ecoflex silicone in a 1:1 ratio by weight and mix thoroughly by hand using a craft stick for three minutes at room temperature (23°C). To fabricate the internal air chambers, the ring mold is designed with separate top and bottom sections. Once the two sections are glued together later in the process, the empty spaces in each section will combine to form the air chambers.

First, the mixture is poured into the bottom section to form the bottom surface and walls. To eliminate air bubbles from the mixture, the entire assembly (mixture and mold) is placed under vacuum at 100 kPa for two minutes. As the mixture begins to solidify after a few minutes, a small hole is punctured in the outer surface to create an opening for the tubing connector used for inflation. The tubing connector is then attached, and additional Ecoflex mixture is used to seal the connection between the tubing and the bottom section. After approximately twenty minutes, the mixture starts to take on a solid form. At this stage, but before it fully cures (which can take several hours), the bottom part is placed onto the base of the TPU casing.

Additional Ecoflex mixture is prepared and poured into the mold for the top section. Vacuum is again used to remove air bubbles. The partially cured bottom section is then positioned directly onto the still-curing top section. This assembly is allowed to cure at room temperature (23°C) for four hours, ensuring a strong bond between the two parts and the proper sealing of the air chambers. By allowing the top section to cure simultaneously with the bottom, we strengthen the mechanical bond between the two surfaces, enhancing the overall durability and performance of the actuator module.

The final stage involves connecting the TPU cap to the base using mechanical screws. Once each actuation module is completed, we stack them in the sequence of compression, longitudinal, compression, and so on, to construct an integrated actuator station. To ensure

alignment among multiple modules, 3D-printed guide rails are used.

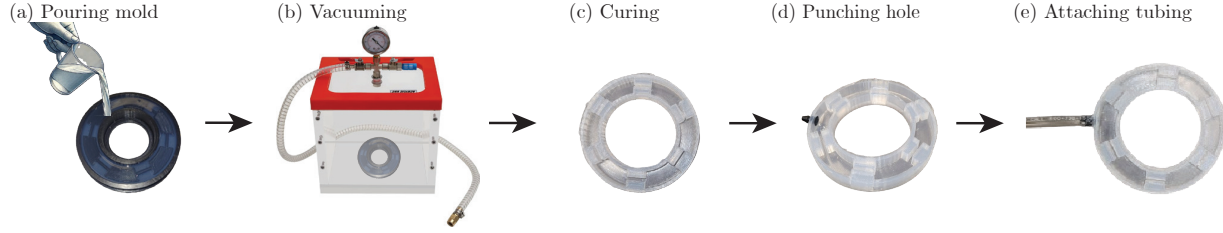


Figure 4.3: Fabrication process of the actuation module. (a) Pouring silicone material into the mold. (b) Removing air bubbles using a vacuum chamber. (c) Allowing the silicone to cure. (d) Creating openings on the outer surface for connectors and tubing. (e) Attaching tubing to finalize the assembly. Steps (a) and (b) are repeated for the upper mold, which is then cured together with the completed bottom mold.

4.3.3 Simulation Framework

The inflation behavior of the actuator was simulated using Finite Element Analysis (FEA) in AbaqusTM (Academic Edition on UCLA Hoffman2 cluster). The simulation results guided the selection of optimal geometric parameters for the experimental setup, helping to minimize material waste, fabrication costs, and time-consuming trial-and-error processes. The final geometric parameters used in the experiments are detailed in [Table 4.1](#), and a more in-depth discussion of the simulation results will be provided in [Section 4.5](#).

The geometry in [Figure 4.2\(c\)](#) was discretized using tetrahedral mesh elements (C3D10), with a 3D model created in SolidWorksTM and imported into AbaqusTM. The mesh included 49,736 nodes and 29,686 elements. A hyperelastic Mooney-Rivlin model was used, with nonlinear geometry enabled in the software. Material properties were approximated for Ecoflex silicone (Shore 00 hardness 45), assigning a Young’s modulus of 100 kPa [\[4\]](#) and a Poisson’s ratio of 0.44 (selected to avoid numerical instability from a Poisson ratio of 0.5 while remaining close to incompressible behavior). The model’s coefficients were $C_{10} = C_{01} = 8620.69$ Pa and $D_1 = 2.4 \times 10^{-6}$ Pa [\[6\]](#). A pressure load of 15 kPa, measured experimentally, was applied to simulate the inflation process. The ratio of the two parameters C_{01} and C_{10} was assumed to be one; we observed that varying this ratio between 0.5 and 10 had a negligible effect on our results.

The boundary conditions were applied differently for the two modules. For the longitudinal module, nodes on the side walls were constrained using an encastre boundary condition, meaning they were fully fixed in place with no translation or rotation allowed, to prevent any deformation. The nodes on the top and bottom walls were free to deform under the applied pressure. In contrast, for the compression module, the nodes on the top and bottom walls, as well as the nodes on the outer surface of the side walls, were constrained. However, the nodes on the inner side walls were free.

For the purposes of numerical simulations, keep in mind that the primary function of the longitudinal module is to provide upward movement through repeated actuation cycles. The total displacement of the object is simply the product of the upward movement per cycle times the number of cycles. Since this is a prototype design study, our primary focus is not on power consumption and efficiency at this stage but rather on feasibility. We can simply increase the number of cycles to achieve the desired movement. In contrast, the compression module is crucial for grasping the object securely during transport. A stronger grasp allows the system to handle a wider variety of objects. Therefore, the focus of our modeling efforts is on refining the geometric parameters of the compression module to maximize its grasping effectiveness. To describe the inflation behavior, we introduce normalized inflation $\bar{d}_c$, defined as the ratio of the maximum radial inflation displacement d_c to the inner radius r of the compression module, as [Figure 4.2\(c\)](#). In simulation, d_c represents the maximum deformation of the nodes positioned along the inner side wall. Additionally, to characterize the object dimension d_o – the maximum cross-sectional dimension – in the context of the system, we use the normalized dimension $\bar{d}_c$, defined as the ratio of d_o to the inner radius r .

4.3.4 Experimental Setup

The overall setup relies on sequential inflation and deflation of the modules to transport an object, as detailed later in [Section 4.4](#). This process requires precise pressure sensing from each actuation module, allowing the system to detect when the object is in contact with the current module through variations in pressure signals. As shown in [Figure 4.4](#), the system

to accomplish this objective includes an air pump and vacuum, air valves, pressure sensors, and a microcontroller, as a closed-loop pneumatic control system.

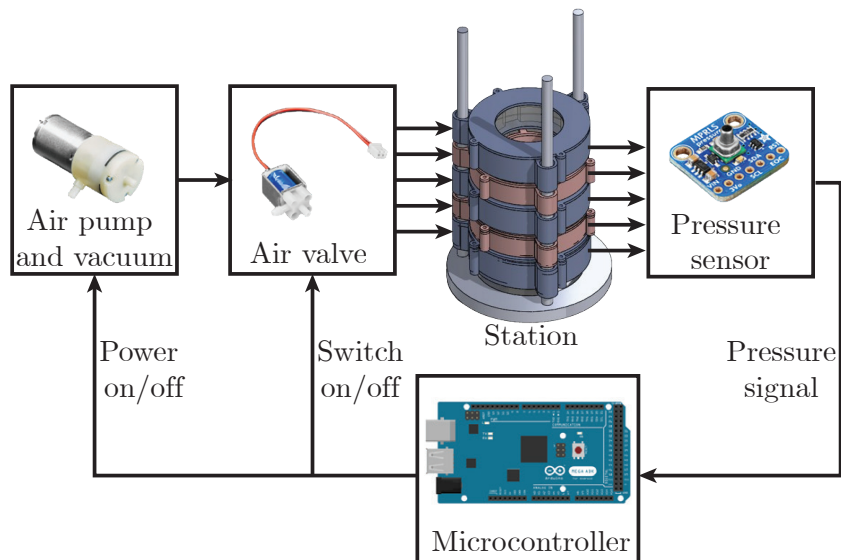


Figure 4.4: Hardware setup. The actuator station incorporates an air pump and vacuum, air valves, pressure sensors, and a microcontroller, providing both sensing and control capabilities.

The pressure sensors (Adafruit™ MPRLS Ported Pressure Sensor Breakout - 0 to 25 PSI) monitor the internal pressures of each actuation module in real time, providing data for feedback loops to control inflation and deflation. This pressure data is transmitted to the microcontroller (Arduino™ Mega 2560 Rev3), which manages airflow supplied by the air pump (Adafruit™ Air Pump and Vacuum DC Motor - 4.5 V and 2.5 LPM - ZR370-02PM).

Following a control sequence outlined later in Section 4.4, the air valves (Adafruit™ 6V Air Valve with 2-pin JST XH Connector - FA0520E) precisely regulate airflow into and out of the individual actuation module.

4.4 Peristaltic Transportation

The peristaltic transportation mechanism in our actuator utilizes sequential inflation and deflation of individual actuation modules to enable controlled movement and object manipulation. This section highlights two key aspects of the system’s operation. First, the

fundamental unit, consisting of three actuation modules, performs basic tasks such as object grasping and transport through sequential control. Second, we examine the concept of synchronization as additional modules are stacked, creating a multi-level structure that necessitates coordinated control across different levels. This coordination ensures consistent and continuous behavior, facilitated by a closed-loop control system.

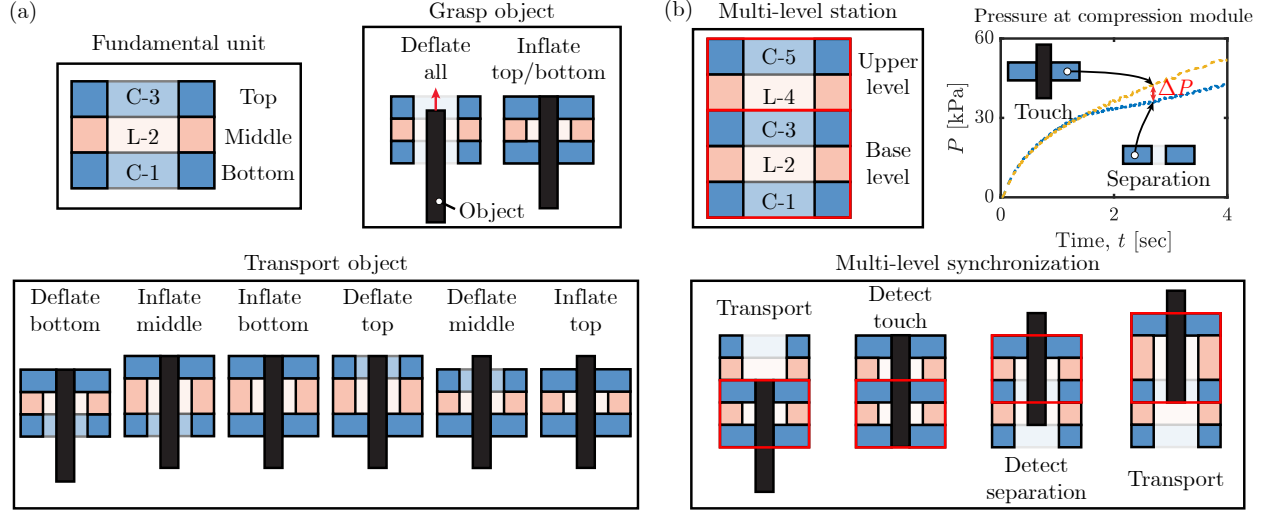


Figure 4.5: Control of peristaltic transportation. (a) The fundamental unit, composed of two compression modules at the top and bottom with a longitudinal module in the middle, enables object grasp and transport through sequence-based control of the actuation modules. (b) The multi-level station, where synchronization is achieved by acquiring pressure signal differences ΔP between the adjacent compression modules. Actuation timing is adjusted based on whether the object is grasped by the upper level modules.

4.4.1 Sequence-Based Control of Fundamental Unit

As shown in Figure 4.5(a), The fundamental unit capable of peristaltic transportation consists of three actuation modules: a compression module at the bottom (C-1), a longitudinal module in the middle (L-2), and another compression module at the top (C-3). This configuration allows for sequential inflation and deflation to achieve object grasp and transport.

To begin, both the top and bottom compression modules are inflated to securely grip the object. Next, to transport the object, the bottom compression module is deflated while the middle longitudinal module inflates, generating an effective axial displacement for the object. Once the middle module has displaced the object, the bottom module is inflated

again to re-grasp, and the top module is deflated. Finally, the middle module is deflated, the top module is re-inflated, and the process is repeated. This cyclical sequence of inflating and deflating the modules ensures that the object is progressively transported through the fundamental unit.

4.4.2 Pressure Feedback Control of Multi-Level Station

As the system is scaled by stacking additional modules – commonly required in practical applications for varying transportation distances – it becomes critical to implement a control strategy that adjusts actuation based on the object’s position along the stacked modules. In the multi-level station, as shown in [Figure 4.5\(b\)](#), the base level functions as a fundamental unit. When additional longitudinal-compression modules (L-4, C-5, etc.) are stacked on top of the base, higher levels are formed. The primary challenge is coordinating the actuation of these upper levels to ensure smooth object transport. Since the object moves from bottom to top, the base level is actuated first. However, determining the optimal timing to activate the upper levels as the object approaches remains a critical problem.

To address this, we utilize pressure signals from sensors integrated into each compression module. When an object contacts the air chamber, the pressure increases more rapidly, exhibiting a higher rate of change compared to when no object is present. During each actuation period, we take multiple pressure measurements at short time intervals. If the pressure change rate exceeds a predefined threshold – established based on inflation without an object – we determine that the current compression module is in contact with the object. Upon detecting contact in an upper compression module, we deactivate the lower module and designate the upper and middle modules as the fundamental unit to transport. By monitoring the pressure change rate in real time, we dynamically adjust the actuation sequence to facilitate smooth and continuous upward transport.

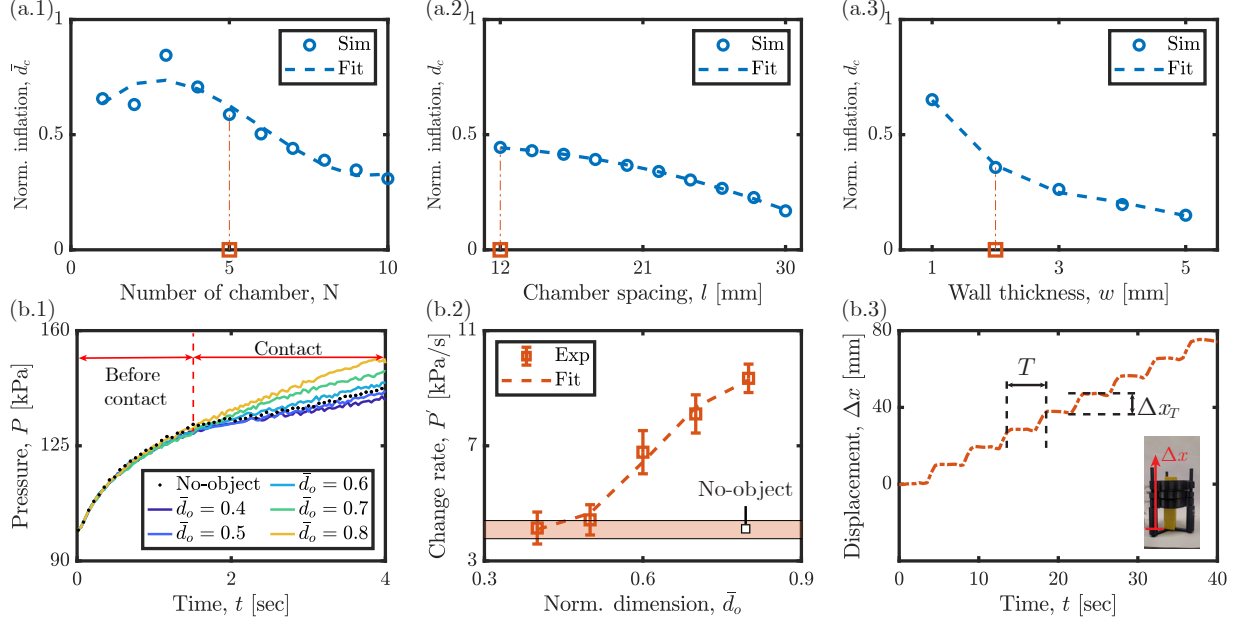


Figure 4.6: Evaluation of the actuator design. (a) Simulation of the geometrical design parameters for the compression module. Blue circles indicate simulation results with a fitted curve. (a.1) The normalized inflation $\bar{d}_c$ plotted against the number of chambers N . (a.2) The normalized inflation $\bar{d}_c$ plotted against the chamber spacing l . (a.3) The normalized inflation $\bar{d}_c$ plotted against the wall thickness w . Red squares highlight the final design parameters selected for the experimental setup: $N = 5$, $l = 12$ mm, and $w = 2$ mm. (b) Experimental evaluation of object grasping and transport. (b.1) Chamber pressure P of compression module over time t when grasping cylinder objects of different normalized diameter $\bar{d}_o$, compared to the no-object case. Each dimension was tested five times, and only one representative curve for each case is shown for clarity. (b.2) Pressure change rate P' of compression module during contact phase, plotted against normalized dimension $\bar{d}_o$. Error bars indicate the mean and variance across multiple experimental trials. (b.3) Object displacement Δx over time t , with the period of the inflation-deflation cycle T corresponding to displacement Δx_T .

4.5 Results and Discussion

In this section, we present simulation results on actuator design and experimentally evaluate grasping and transportation performance. First, we use finite element analysis (FEA) to explore the geometric parameter space of the compression module. Based on these findings, we determine the optimal parameters for the experimental setup to achieve effective actuation. We then conduct inflation tests to characterize object transport and assess the range of object sizes suitable for grasping and movement. Finally, we examine the pressure-based multi-level synchronization strategy and demonstrate how different module sequences impact

transportation speed.

4.5.1 Parameter Space for Actuation Module Design

The compression module plays a critical role in object grasping. In [Figure 4.2\(c\)](#), we use the radial inflation displacement, d_c , to characterize the range of object dimension d_o that can be grasped, i.e., $d_o/2 \in (r - d_c, r)$. Assisted by FEA, we investigate how geometric parameters of the ring influence the radial inflation displacement. Detailed simulation results can be found in the supplementary video.

To comprehensively describe the geometry of the compression ring, we consider seven key parameters, as illustrated in [Figure 4.2\(a\)](#). The step height m and chamber width h define the dimensions of the air tunnel and chambers in the direction perpendicular to the airflow, affecting how efficiently air can inflate between the chambers. However, given the already small system dimensions, there is limited flexibility in adjusting these parameters without risking airflow impedance. As a result, we exclude these parameters from further investigation, selecting values that balance ease of fabrication with effective airflow. While the outer radius R and inner radius r define the critical dimensions, we study how the remaining parameters impact the inflation behavior, characterized by the normalized inflation $\bar{d}_c$.

The variables that define the air chamber distribution around the ring include the chamber length l , chamber spacing s , and the number of air chambers N . These parameters are interrelated and governed by the following relationship

$$(s + l) \times N = \pi(R + r).$$

Therefore, we investigate how variations in N and l influence inflation, with s coupled to these parameters while R and r remain fixed. As shown in [Figure 4.6\(a.1\)](#), we observe that $\bar{d}_c$ does not monotonically change with N . When $N = 1$ or 2 , $\bar{d}_c$ is lower than for $N = 3$ due to severe self-contact as discussed before. As the number of chambers increases, $\bar{d}_c$ decreases because the increased air chamber volume inflates less under a constant pressure

loading. Additionally, as shown in Figure 4.6(a.2), we observe that as the spacing l increases, $\bar{d}_c$ decreases. This is because chambers that are spaced too sparsely inflate more slowly, resulting in less efficient inflation-deflation cycles.

Furthermore, wall thickness is essential in determining surface stiffness. Thinner walls inflate more easily, as shown in Figure 4.6(a.3), but they are also more susceptible to damage, such as bulging or bursting under pressure.

To balance between structural robustness and inflation performance, we selected a set of appropriate geometric parameters, marked as red squares in Figure 4.6(a), with $s = 28$ mm, $l = 12$ mm (corresponding to $N = 5$), and $w = 2$ mm. These parameters were used consistently across all the following experiments.

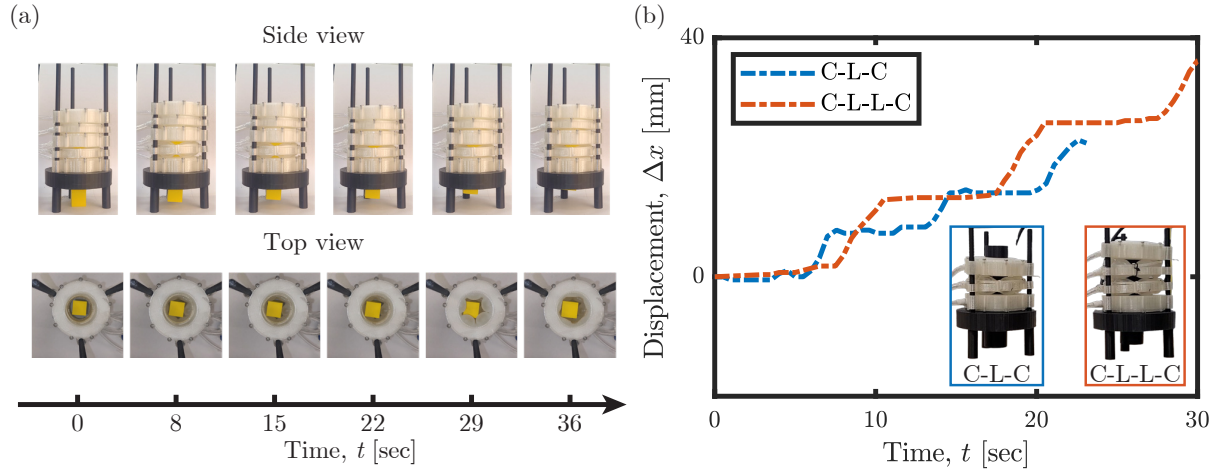


Figure 4.7: Demonstration of the modular features on system variation. (a) Multi-level synchronization in object transportation. (b) Module sequence on object transportation performance during three periods, i.e., C-L-C of 1.00 ± 0.11 mm/s and C-L-L-C of 1.23 ± 0.06 mm/s.

4.5.2 Object Grasping and Transportation

For effective grasping, the object dimension d_o must not exceed the inner diameter of the actuation ring. If d_o is too small, the inflated chamber of the compression module may not make sufficient contact. Conversely, if d_o is too close to the inner diameter, contact is ensured, but inflation may be restricted, potentially damaging the air chamber.

To determine the effective range of object dimensions, we use cylinders with diameter d_o , ranging from 0.4 to 0.8 times the inner diameter of the actuation ring. As shown in [Figure 4.6\(b.1\)](#), upon inflation, we observe an initial transition phase – termed the “before contact” stage – lasting about 1.5 seconds, during which the pressure curves follow similar trends regardless of the object’s dimensions. Following this, in the “contact” stage, larger dimensions result in a more rapid pressure change rate due to increased squeezing.

To facilitate comparison, we compute the slope of each pressure curve during the “contact” stage, represented as the pressure change rate P' . In [Figure 4.6\(b.2\)](#), we present P' for varying cylinder dimensions d_o . Each dimension was tested five times to enhance result reliability, with error bars indicating variability. The baseline, shown as the orange band, is established with no object present. We find that when $\bar{d}_o = 0.4$, P' falls within the band zone, indicating that the compression module cannot reliably detect the presence of an object through pressure change rate, although it may still grasp it. As a result, the effectiveness of multi-level synchronization through pressure feedback would be limited. However, this minimal object dimension may vary depending on different module design parameters.

Additionally, to evaluate the transportation performance, we use a fundamental unit to move a soft cuboid with $\bar{d}_o = 0.708$ (squared cross-section with side length of 20 mm) and mass of $m = 5$ g and track its displacement Δx over time. Due to the nature of sequential control, the object only moves when the longitudinal module is inflated. As shown in [Figure 4.6\(b.1\)](#), the object exhibits periodic upward movement, characterized by a period $T = 4.5$ s (representing the time of an inflation-deflation cycle of the fundamental unit) and a corresponding displacement $\Delta x_T = 9.5$ mm (the inflation distance of the longitudinal module). The average transportation speed is simply $v_o = \Delta x_T / \Delta T \approx 2.08 \pm 0.07$ mm/s.

4.5.3 Pressure-Based Synchronization

Leveraging the advantage of modular design, the modules can be easily stacked in varying lengths to form a multi-level station, allowing for the transportation of objects over different distances. Using a pressure-based control strategy, we demonstrate its ability to transport

objects smoothly and continuously across different levels.

As shown in [Figure 4.7\(a\)](#), we use a two-level station to transport the soft cuboid with $\bar{d}_o = 0.708$. At $t = 0$, the object is in contact with compression modules C-1 and C-3. Once C-3 inflates, we also inflate C-5 and monitor its chamber pressure. The inflation-deflation cycle period, T , for the fundamental units is 4.5 s. During the first four cycles from $t = 0$ to $t = 18$ s, the pressure in C-5 remains similar to its no-object condition, with $P' \approx 4.33$ kPa/s. The upper longitudinal module L-4 remains inactive, and the fundamental unit consisting of C-1, L-2, and C-3 continues to operate. However, at $t = 22$ s, the pressure in C-5 suddenly increases to 8.48 kPa/s, approximately doubling, signaling the detection of the object. The system then activates the upper level, which includes L-4 and C-5, to work in tandem with C-3, while deactivating C-1 and L-2. This process continues, activating additional levels in sequence until the object reaches the top. This approach ensures smooth, synchronized transport across the stacked modules. Further details of the transportation process can be found in the supplementary video.

4.5.4 Module Sequence on Transportation Speed

As an another benefit of the modular design, the system allows for varying actuation module sequences. The fundamental units consist of two compression modules and one longitudinal module for transportation. However, by inserting additional longitudinal modules, we can study how the module sequence affects transportation speed.

As shown in [Figure 4.7\(b\)](#), compared with the conventional C-L-C configuration, the C-L-L-C configuration changes the period T (from 9 s to 13 s) and corresponding displacement Δx_T (from 7.5 mm to 13 mm). On one hand, T increases, as we use one pump to inflate two longitudinal modules, which consumes the similar energy. Although the inflation time of C-L-L-C configuration is longer, it does not double as C-L-C configuration, only increasing by about 1.5 times. On the other hand, Δx_T for C-L-L-C configuration almost doubles, since both longitudinal modules contribute to displacement. Overall, the transportation speed of C-L-L-C configuration increases by 1.2 times compared to the C-L-C configuration,

achieving higher speed by simply reordering the actuation module sequence. Further details on the configuration comparison are available in the supplementary video.

4.6 Conclusions

In conclusion, this study presents a bio-inspired pneumatic modular actuator designed for peristaltic transport, highlighting its potential for object manipulation of varying shape and size. By developing distinct actuation modules – compression and longitudinal by either stack number or sequence – we have established a scalable, adaptable architecture that allows for varying transportation tasks. The integration of pressure feedback mechanisms enhances the system’s ability to detect object position.

Future work will focus on systematically exploring the parameter space of both longitudinal and compression modules while optimizing efficiency through adjustments to the actuation period and configuration sequence. Additionally, efforts will be made to seamlessly integrate the system with existing robotic platforms, such as manipulators, to handle fragile objects like coral reefs and strawberries, thereby expanding its practical applications.

CHAPTER 5

Pneumatic Membrane Shells for Robotic Hemostasis

5.1 Motivation

Endovascular procedures have become a cornerstone of contemporary medicine, enabling minimally invasive treatment of a wide spectrum of vascular pathologies, including percutaneous coronary interventions (PCI) and neurointerventional thrombectomies [63], [64]. These techniques reduce patient trauma, accelerate recovery, and lower complication rates compared to open surgery. Despite these benefits, achieving reliable hemostasis at vascular access sites—most commonly the femoral artery—remains a persistent and critical challenge following catheter removal.

Current clinical practice relies on manual compression or semi-passive devices such as the *FemoStop* [65], [66]. Manual compression requires 15–60 minutes of sustained force application, which is inherently prone to operator variability, fatigue, and inconsistency [67], [68]. These limitations frequently result in delayed hemostasis, re-bleeding, or hematoma formation. Major femoral bleeding complications occur in 3–6% of PCI cases, with in-hospital mortality approaching 3% [69], [70]. Up to 10% of patients develop access-site hematomas, extending hospitalization and adding \$10,000–\$15,000 per case [71]–[73]. Such complications increase transfusion rates, ICU admissions, and overall healthcare costs [74].

The central research gap lies in the inability of existing hemostasis methods to deliver adaptive, precise, and continuous pressure control tailored to patient-specific anatomy and physiology. Manual methods lack endurance and precision, while current mechanical devices remain passive, requiring manual oversight and providing no real-time feedback. Critically, these approaches cannot detect subsurface bleeding or track vessel shifts due to patient move-

ment. Addressing this unmet need requires intelligent, feedback-driven systems capable of autonomously monitoring and regulating the biomechanical interactions at the tissue–device interface.

This work seeks to address these limitations by developing a robotic hemostasis system that integrates advanced sensing modalities with closed-loop control. By moving beyond scalar pressure readings to include full-field visuo-tactile deformation mapping and continuous ultrasound imaging, the platform enables unprecedented “contact awareness.” This multi-modal sensing approach allows the system to dynamically localize vessels, detect bleeding, and precisely modulate compression, thereby reducing complication rates and alleviating clinician workload [3], [56].

5.2 Introduction

Endovascular interventions, including percutaneous coronary interventions (PCI) and neurointerventional thrombectomies, are indispensable in cardiovascular and neurovascular care [63], [64]. These minimally invasive techniques have transformed treatment by reducing trauma, shortening recovery times, and improving patient outcomes compared to open surgical approaches. Despite these advances, femoral artery access remains a common route for catheterization, used in 40–60% of PCI cases [75]–[77]. While clinically effective, transfemoral access carries inherent risks: inadequate hemostasis at the puncture site is one of the most frequent and consequential complications. Major access-site bleeding occurs in 3–6% of cases and hematoma formation affects up to 10% of patients [69]–[72]. These complications extend hospital stays, increase costs by \$10,000–\$15,000 per patient, and are associated with elevated in-hospital mortality [73], [74]. Collectively, femoral-access complications impose a substantial burden on patients, providers, and healthcare systems.

Conventional methods of hemostasis are limited. Manual compression, though widely practiced, requires sustained force application for 15–60 minutes and is inherently inconsistent due to operator fatigue, variability in technique, and patient-specific factors such as obesity or tissue compliance [67], [68]. Semi-passive mechanical compression devices, such as

the *FemoStop*, can reduce clinician effort but remain passive and rely on intermittent manual adjustment [65], [66]. Neither approach provides real-time sensing or adaptive feedback, leaving clinicians uncertain about whether the vessel has fully occluded, whether bleeding persists beneath the skin, or whether excessive pressure risks tissue damage. The absence of intelligent, automated control underscores an urgent need for next-generation solutions capable of delivering consistent, safe, and efficient hemostasis.

To address this gap, we propose a fully integrated robotic hemostasis platform (Fig. 4.1) that combines soft pneumatic actuation with co-located, multi-modal sensing. The system incorporates four complementary modalities: robotic arm force feedback, internal membrane pressure sensing, high-speed visuo-tactile deformation mapping, and continuous ultrasound imaging. These redundant sensors enable closed-loop adaptive control, allowing the robot to dynamically detect bleeding, localize vessels, and precisely adjust compression in real time. In contrast to traditional scalar force or pressure measurements, the visuo-tactile module reconstructs full-field displacement maps across the contact surface, providing high-resolution insights into tissue deformation and strain. Meanwhile, a stretchable ultrasound array embedded within the pneumatic membrane provides continuous subsurface monitoring of vessel patency and blood flow [56]. Together, these sensing modalities grant the system unprecedented situational awareness of both surface-level and subsurface tissue mechanics, enabling safe, adaptive, and standardized hemostasis.

This paper details the design, modeling, and validation of the proposed robotic platform. First, we describe the mechanical design of the soft pneumatic end-effector and the integration of the multi-modal sensing suite. Second, we present computational and experimental methods, including finite element analysis, phantom testing, and preliminary *in vivo* swine studies, to evaluate device performance. By benchmarking against current clinical standards, this work demonstrates the potential of robotic, feedback-driven hemostasis to reduce complications, improve workflow efficiency, and alleviate clinician workload. Finally, given the projected growth of the vascular closure device market [78]–[80], the proposed system represents not only a technical advance but also a timely opportunity for clinical translation and commercialization.

5.3 Methods

5.3.1 Pneumatic-Driven Membrane

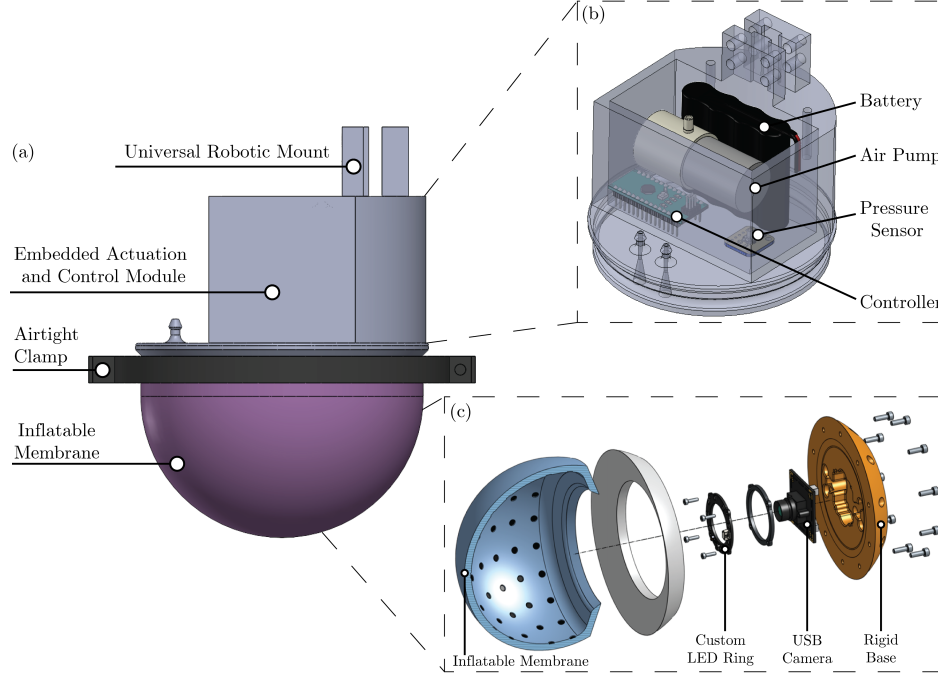


Figure 5.1: Design schematic of the self-contained robotic hemostasis end-effector. (a) Main assembly showing the universal robotic mount for integration with robotic manipulators, an embedded actuation and control module, an airtight clamp ensuring sealed pressurization, and a compliant inflatable membrane for direct tissue contact. (b) Pneumatic subsystem, illustrating internal components including a rechargeable battery, miniature air pump for membrane inflation, pressure sensor for closed-loop regulation, and onboard controller enabling autonomous actuation. (c) Exploded view of the integrated visual-tactile sensing unit, composed of an inflatable membrane with embedded perforations, a custom LED illumination ring, a USB camera for imaging, and a rigid base with fasteners for structural support and modular assembly.

The architecture of the proposed robotic hemostasis system is illustrated in [Figure 5.1](#). The device is designed as a fully self-contained, pneumatically actuated end-effector that enables precise and compliant interaction with patient anatomy. Unlike conventional pneumatic systems that rely on bulky compressors and external tubing, our end-effector integrates all required components into a compact housing measuring approximately $6 \times 6 \times 3 \text{ in}^3$ ([Figure 5.1\(a\)](#)). This miniaturization is achieved by embedding a rechargeable battery, a miniature air pump, a precision pressure sensor, and an onboard controller into a single actuation

and control module. By eliminating external wiring and air lines, the device reduces surgical clutter and facilitates rapid deployment in clinical environments.

The system is designed for compatibility with standard collaborative robotic platforms. A universal robotic mount (Figure 5.1(a)) provides stable attachment, ensuring both ease of integration and reliable positioning during sterile procedures. At the patient interface, a compliant inflatable membrane fabricated from medical-grade silicone (Elite Double 16) provides direct tissue contact. The soft surface conforms continuously to the skin, distributing forces evenly and reducing localized stress concentrations that can occur with rigid end-effectors. This skin-compliant interaction is crucial for minimizing patient discomfort and avoiding tissue injury.

The pneumatic subsystem is depicted in Figure 5.1(b). Internal pressure is regulated by a lightweight onboard controller operating a closed-loop feedback system. A two-way valve and miniature air pump work together to stabilize the applied compression at a desired setpoint: if the sensed pressure falls below the target, the pump inflates the membrane; if it exceeds the threshold, the valve opens to release air. This closed-loop strategy enables adaptive adjustments in response to variations in patient tissue compliance or changes in robotic arm positioning, ensuring safe and effective hemostasis.

Finally, Figure 5.1(c) shows an exploded view of the integrated visuo-tactile sensing unit. This assembly combines the inflatable membrane with an internal perforated marker layer, a custom LED illumination ring, and a miniature USB camera. The rigid base provides structural support while allowing modular assembly. This design not only enables full-field deformation tracking but also leaves flexibility for incorporating auxiliary diagnostic tools, such as ultrasound or Doppler probes, without compromising sterility or compactness.

Together, the elements shown in Figure 5.1 demonstrate how the pneumatic-driven membrane is engineered to balance portability, compliance, and precise control—qualities that make it well-suited for clinical deployment in robotic hemostasis.

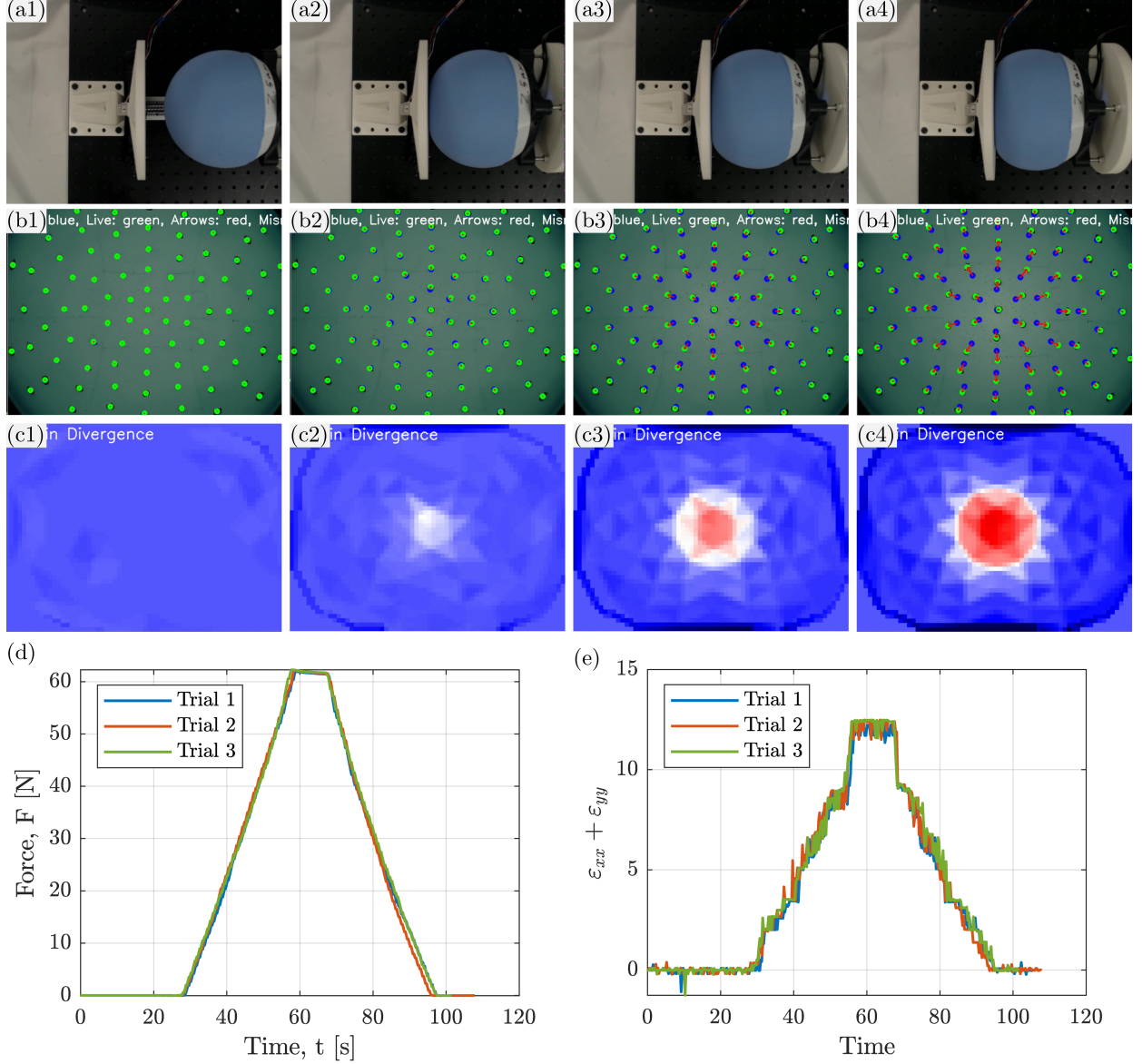


Figure 5.2: Characterization of the visual–tactile sensing process. (a) Top-view snapshots of the inflatable chamber as it indents against a load cell at increasing distances. (b) Camera observations of the marker field during deformation. Blue dots indicate the initial (undeformed) marker positions, green dots indicate real-time positions under deformation, and red arrows represent displacement vectors from the initial to the deformed states, visualizing the local strain field. (c) Strain divergence maps computed from the displacement field in (b). Divergence quantifies the local areal expansion or contraction of the marker field, with red regions indicating strong outward expansion and blue regions indicating negligible change. (d) Measured contact force F as a function of time across three repeated trials, showing consistent loading and unloading cycles. (e) Evolution of the spatially averaged strain component ($\epsilon_{xx} + \epsilon_{yy}$) over time for the same trials, demonstrating strong agreement with the measured contact force and validating the reliability of the visual–tactile sensing approach.

5.3.2 Multi-Modal Sensor Fusion

Precise, adaptive, and autonomous hemostasis requires more than scalar force or pressure measurements. Our system therefore integrates a comprehensive multi-modal sensing architecture (Figure 5.2) that combines robotic arm force feedback, internal membrane pressure sensing, high-speed visuo-tactile deformation mapping, and continuous ultrasound imaging. By fusing these complementary modalities, the system achieves robust situational awareness of both surface-level tissue mechanics and subsurface vascular status, enabling closed-loop control.

The pneumatic end-effector inherently provides continuous internal pressure readings, while the robotic arm supplies global force and torque feedback. These global signals, however, cannot capture localized strain distributions at the tissue-device interface. To address this, we embed a visuo-tactile sensor directly within the inflatable membrane, allowing for high-resolution monitoring of full-field surface deformation. As shown in Figure 5.2(a), the chamber is indented against a load cell while its internal marker field is imaged in real time. Figure 5.2(b) illustrates the sensing principle: each green dot represents the deformed position of a marker, while blue dots denote the reference configuration. Red arrows indicate displacement vectors $\mathbf{u} = (u_x, u_y)$ between initial and deformed states.

From these displacements, we compute the two-dimensional strain tensor components using finite-difference approximations of displacement gradients:

$$\varepsilon_{xx} = \frac{\partial u_x}{\partial x}, \quad \varepsilon_{yy} = \frac{\partial u_y}{\partial y}, \quad \varepsilon_{xy} = \frac{1}{2} \left(\frac{\partial u_x}{\partial y} + \frac{\partial u_y}{\partial x} \right). \quad (5.1)$$

The local areal strain, or divergence, is then given by

$$\nabla \cdot \mathbf{u} = \varepsilon_{xx} + \varepsilon_{yy}, \quad (5.2)$$

which quantifies the local expansion or contraction of the marker field. Figure 5.2(c) shows divergence maps derived from the displacement field, where red regions indicate outward expansion and blue regions denote minimal strain.

Force measurements recorded by the load cell (Figure 5.2(d)) exhibit a strong correlation with the spatially averaged areal strain $\langle \varepsilon_{xx} + \varepsilon_{yy} \rangle$ (Figure 5.2(e)). Across repeated trials, both force F and averaged strain evolve consistently during loading and unloading cycles, validating that strain divergence provides a reliable surrogate for contact force. This relationship enables the robot to infer distributed forces directly from visuo-tactile feedback without relying solely on global load-cell measurements.

The integration of ultrasound further complements these measurements by providing subsurface information. Embedded within the membrane, a stretchable ultrasound array continuously monitors vessel patency and blood flow. This unique co-location of tactile and ultrasound sensing ensures that both surface and subsurface phenomena are captured at the precise site of compression. For example, if tactile feedback detects localized overstress while ultrasound simultaneously reveals residual flow, the control logic can adjust inflation pressure to maintain hemostasis while minimizing tissue injury.

Finally, data from all four modalities—robotic arm force/torque, membrane pressure, visuo-tactile strain fields, and ultrasound imaging—are fused through a high-frequency multi-layered control framework (exceeding 10 Hz). A Kalman filter continuously integrates these signals to estimate the membrane-tissue contact state, while rule-based logic governs pressure adjustments. Specifically, if ultrasound detects ongoing perfusion, inflation pressure is increased; conversely, if tactile strain divergence exceeds safe thresholds, pressure is reduced. Through this strategy, the system autonomously localizes the femoral access site, dynamically regulates compression, detects bleeding, and ensures stable and effective closure.

Together, the experimental results in Figure 5.2 demonstrate the validity of using visuo-tactile strain divergence as a proxy for contact force, and the necessity of multi-modal fusion for achieving safe, adaptive robotic hemostasis.

5.3.3 Integrated System

The performance of the pneumatic-driven membrane end-effector was experimentally characterized to validate its ability to deliver controlled, distributed compression (Figure 5.3).

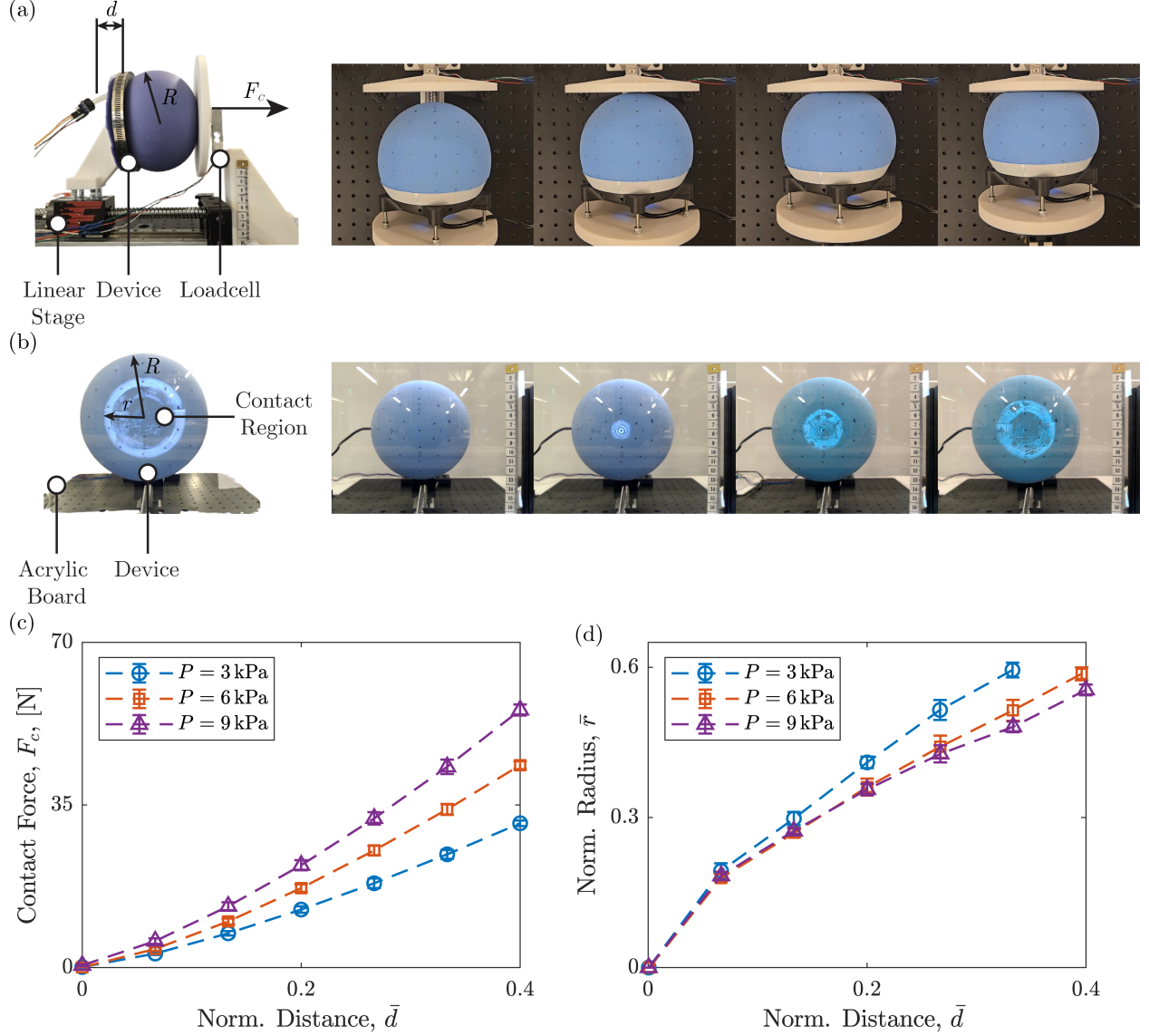


Figure 5.3: Experimental characterization of the inflatable membrane end-effector. (a) Side-view setup for contact force measurement. The device is mounted on a linear stage that advances a prescribed distance d , pressing the inflated membrane of radius R against a load cell to measure the resulting contact force F_c . Sequential images illustrate membrane deformation at increasing indentation depths. (b) Lateral-view setup for contact area characterization. The device is pressed against a transparent acrylic board while a camera captures the contact region. The inflated membrane of radius R produces a near-circular contact area characterized by radius r , with images showing the progression of contact growth under increasing compression. (c) Measured contact force F_c as a function of normalized indentation distance $\bar{d}$ for chamber pressures of 3, 6, and 9 kPa, showing a monotonic increase in force with both displacement and pressure. (d) Normalized contact radius $\bar{r}$ as a function of normalized indentation distance $\bar{d}$ under the same pressure conditions, demonstrating consistent growth of the contact region with displacement and a dependence on internal pressure.

These experiments evaluated both the force–displacement response and the growth of the contact region under varying inflation pressures, providing key insights into the device’s interaction with soft tissue.

Figure 5.3(a) illustrates the side-view experimental setup for measuring contact force. The device was mounted on a linear stage and advanced by a prescribed indentation distance d against a load cell. The chamber radius R was kept constant during inflation, and sequential snapshots show the deformation of the inflated membrane as indentation increased. This setup enabled direct quantification of the contact force F_c transmitted through the membrane at controlled displacements.

To assess the distribution of contact, a lateral-view experiment was conducted as shown in Figure 5.3(b). Here, the device was pressed against a transparent acrylic board while a camera captured the resulting contact region. The inflated membrane produced a near-circular contact area characterized by radius r , which expanded with increasing indentation. Visual observations confirm that the compliant silicone membrane deforms smoothly and maintains consistent geometry, which is critical for ensuring uniform pressure delivery to patient tissue.

The results of these characterizations are summarized in Figure 5.3(c) and Figure 5.3(d). When the normalized indentation depth $\bar{d} = d/R$ was increased, the measured contact force F_c grew monotonically under all tested chamber pressures (3, 6, and 9 kPa). As expected, higher inflation pressures amplified the force response, reflecting the nonlinear stiffness contributed by membrane inflation (Figure 5.3(c)). Simultaneously, the normalized contact radius $\bar{r} = r/R$ increased consistently with $\bar{d}$ (Figure 5.3(d)), demonstrating that deeper indentation and higher pressures enlarge the effective contact area. This scaling behavior is consistent with thin-shell membrane mechanics, where force is transmitted through both normal indentation and circumferential membrane tension.

Together, these experiments establish the controllable and predictable mechanics of the integrated system. The ability to tune both the magnitude of compression (via inflation pressure) and the distribution of contact (via membrane deformation) is essential for clinical

safety: sufficient force must be applied to occlude arterial flow, while excessive localized pressure must be avoided to prevent tissue damage. The characterization in Figure 5.3 confirms that the inflatable membrane achieves this balance, providing a quantitative foundation for closed-loop hemostasis control.

5.3.4 Angle-Dependent Strain-to-Force Mapping

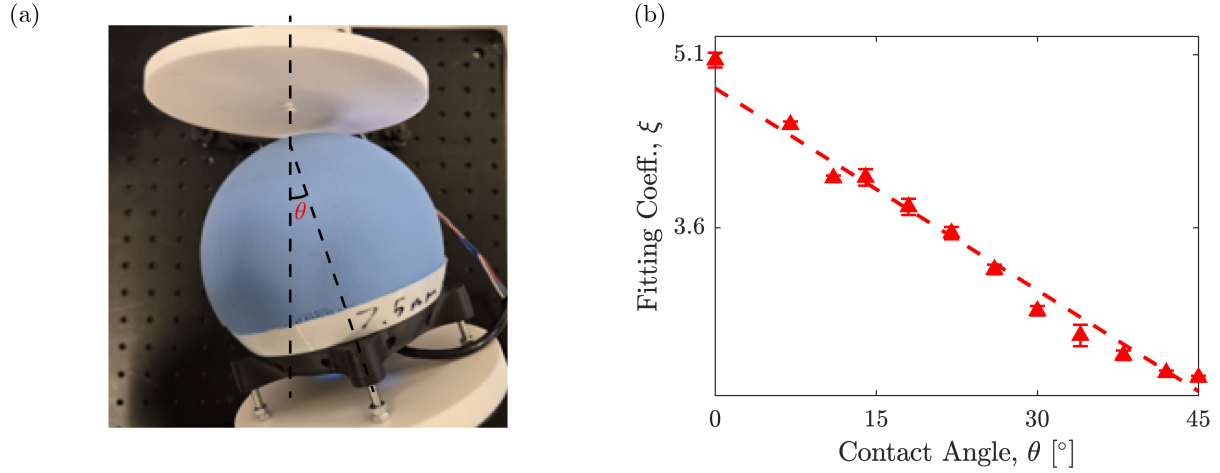


Figure 5.4: Effect of contact angle on the calibration between strain and force. (a) Experimental setup showing the inflatable membrane indented by a flat plate at varying approach angles θ relative to the membrane axis. (b) Fitting coefficient ξ that maps spatially averaged strain divergence ($\varepsilon_{xx} + \varepsilon_{yy}$) to measured force, plotted as a function of θ from 0° to 45° . The decrease in ξ with increasing angle arises from geometric effects and optical distortion in the camera view, highlighting the need to incorporate angular dependence into calibration.

The visuo-tactile sensing framework relies on a calibration that links measured strain divergence ($\varepsilon_{xx} + \varepsilon_{yy}$) to applied contact force F . This relationship is typically expressed as

$$F \approx \xi \langle \varepsilon_{xx} + \varepsilon_{yy} \rangle, \quad (5.3)$$

where ξ is a fitting coefficient obtained from experimental calibration, and the angle brackets denote spatial averaging across the contact region.

However, when the contact interface is not perfectly aligned with the camera's optical axis, distortions alter the apparent displacement field. To systematically study this effect, we varied the contact angle θ between the indenter plate and the axis of the inflated membrane

(Figure 5.4(a)). Across a range of $0^\circ \leq \theta \leq 45^\circ$, we extracted the corresponding fitting coefficient ξ by regressing the measured force against the computed strain divergence.

The results in Figure 5.4(b) show a monotonic decrease in ξ as θ increases, with values dropping from approximately 5.1 at normal incidence ($\theta = 0^\circ$) to below 3.0 at $\theta = 45^\circ$. This trend reflects two main contributions: (i) geometric projection errors, where off-axis contact reduces the effective strain captured by the camera, and (ii) optical distortions introduced by the spherical membrane surface at oblique viewing angles. Together, these effects lead to underestimation of the true strain field when viewed at larger θ .

Accounting for this angular dependence is essential for accurate strain-to-force mapping in practical use cases, where the robot may not always compress the membrane at normal incidence. By incorporating θ as a correction factor in calibration, the system can maintain reliable force inference across different approach orientations. This adjustment strengthens the robustness of the multi-modal sensing framework and ensures consistent performance during in situ robotic hemostasis.

5.3.5 Surrogate Model for Gap Closure Mechanics

To better understand the mechanics of vascular access site closure, we designed a surrogate experimental model inspired by the process of hemostasis. In clinical scenarios, compression is applied externally on the skin and transmitted through soft tissue to collapse the puncture gap surrounding a bleeding vessel. We replicated this scenario with a silicone block embedding a vessel-like cavity and a pre-cut channel representing the separation between tissue flaps (Figure 5.5(a)–(b)). The inflatable membrane applies distributed pressure from above, mimicking manual or robotic compression in practice.

The geometry of the surrogate model is parameterized by the block dimensions (L , H), the vessel diameter D , the channel depth d , and most importantly, the initial gap width l (Figure 5.5(c)). The gap l serves as the dominant factor controlling whether complete closure is achievable. During compression, l progressively decreases until closure is reached or until the applied load becomes insufficient to overcome tissue resistance.

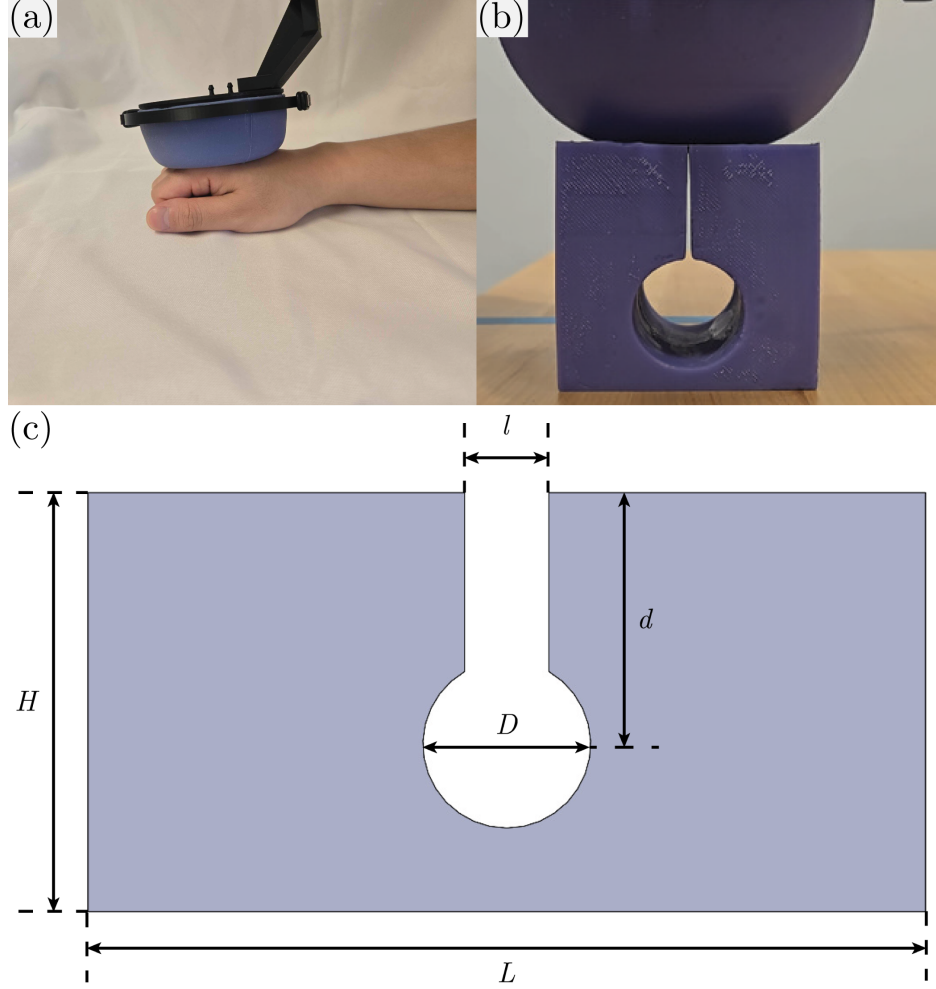


Figure 5.5: Analogy-based surrogate model for hemostasis and characterization of gap closure under compression. (a) Conceptual translation from clinical hemostasis: the inflatable membrane applies compression on skin tissue, here mimicked by a silicone surrogate block. (b) Experimental surrogate model enclosing a mock vessel. A pre-cut channel with initial gap l represents the separation between tissue flaps covering the vessel. (c) Geometric definition of the surrogate model, characterized by block width L , height H , channel depth d , and vessel diameter D . The initial gap l is the critical parameter governing closure. (d) Comparison of finite element analysis (FEA) snapshots and real experimental images during loading, showing progressive narrowing of the gap as pressure increases. (e) Evolution of the gap size under pressure loading for initial gaps ranging from $l = 2$ mm to 10 mm. Narrow gaps (2–4 mm) close rapidly under moderate pressure, while large gaps (e.g., 10 mm) cannot be fully closed due to excessive separation. These results demonstrate that hemostasis efficacy depends strongly on the initial tissue gap.

To capture the mechanics of this process, we combined finite element analysis (FEA) with physical experiments. Representative snapshots from FEA simulations and real tests

(Figure 5.5(d)) show strong qualitative agreement: under increasing loading, the gap narrows smoothly and eventually pinches closed for small initial gaps. The experimental data further confirm that closure is highly sensitive to l .

Quantitative results are summarized in Figure 5.5(e), which plots the evolution of gap size as a function of applied pressure for different initial values of l (2–10 mm). Small gaps ($l = 2$ mm) collapse rapidly under modest pressure, achieving closure efficiently. Intermediate gaps (4–6 mm) require higher pressures but remain closable within the tested range. In contrast, large gaps ($l = 10$ mm) fail to close completely even under maximum applied loading, indicating a geometric limitation inherent to the system.

These findings emphasize that successful hemostasis depends not only on applied pressure but also on the initial tissue configuration. Clinically, this suggests that closure devices or robotic systems must adapt to patient-specific anatomy, particularly when the tissue gap is large due to vessel dilation, repeated punctures, or tissue damage. By validating the surrogate model through both simulation and experiment, we establish a framework for predicting closure performance and informing design guidelines for adaptive robotic compression strategies.

5.4 Conclusions

This work presented the design, modeling, and experimental validation of a robotic hemostasis platform that addresses longstanding limitations in vascular access site closure. By integrating a pneumatically driven soft membrane with multi-modal sensing—including robotic force feedback, internal pressure measurement, visuo-tactile strain mapping, and real-time ultrasound imaging—the system achieves adaptive, closed-loop control capable of delivering consistent and safe compression.

Experimental characterizations demonstrated that the inflatable membrane provides tunable force-displacement responses and predictable contact area growth under loading, critical for distributing pressure evenly across patient tissue. Visuo-tactile sensing was shown to capture full-field strain divergence that correlates strongly with measured force, while

ultrasound sensing provided continuous subsurface feedback on vessel patency. Together, these modalities established a comprehensive sensing architecture that enables the robot to dynamically detect bleeding, localize vessels, and prevent tissue over-compression. Surrogate models and finite element analysis further highlighted how anatomical factors, such as initial tissue gap size, govern closure efficacy, emphasizing the importance of adaptive pressure regulation.

The results suggest that robotic hemostasis can substantially improve clinical reliability compared to manual compression or passive devices, reducing complications such as hematoma and access-site bleeding while alleviating clinician workload. Moreover, the platform provides a versatile foundation for broader applications requiring safe, feedback-driven compression, ranging from wound care to diagnostic palpation.

Future work will focus on translational studies, including in vivo validation, extended safety assessments, and integration with clinical workflows. By bridging advances in soft robotics, multi-modal sensing, and intelligent control, this platform represents a significant step toward standardized, automated hemostasis and opens new opportunities for robotic systems in interventional medicine.

CHAPTER 6

Conclusions

This dissertation has explored the fundamental mechanics, design strategies, and biomedical potential of soft robotic systems through three canonical archetypes: rods, rings, and shells. By grounding the investigation in biologically inspired principles and advancing both modeling and experimental validation, this work provides a unified framework for understanding how compliant structures can be harnessed for safe, adaptive, and multifunctional robotic platforms.

6.1 Summary of Contributions

The first major contribution lies in the study of **rod-like systems**, which emulate flagellar propulsion and slender-body locomotion. Through the development of discrete elastic rod models and coupled fluid–structure simulations, this work has shown how geometric nonlinearities, buckling instabilities, and hydrodynamic interactions can be leveraged to generate thrust and maneuverability. These insights extend beyond low-Reynolds-number swimming, offering pathways for steerable catheters, soft continuum manipulators, and filament-based navigation in biomedical environments.

The second contribution focuses on **ring-based architectures** inspired by peristaltic transport. By designing and characterizing multi-segmented soft actuators, this research demonstrates how sequential activation and anchoring mechanics can enable efficient locomotion and transport in confined spaces. Experimental validation with modular ring systems highlights their robustness, adaptability to variable geometries, and scalability across applications ranging from capsule endoscopy to extraterrestrial material handling. The work

bridges biological peristalsis with robotic design, offering a blueprint for rhythmic transport mechanisms that balance simplicity with effectiveness.

The third contribution centers on **shell-inspired actuation**, emphasizing hyperelastic membranes for distributed surface interaction. By integrating visuo-tactile sensing, pneumatic control, and ultrasound monitoring, this research establishes a robotic platform capable of adaptive compression, vessel localization, and bleeding detection. Beyond its direct application to vascular hemostasis, the framework underscores how soft shells can serve as versatile interfaces for conformal grasping, tactile mapping, and multifunctional biomedical interventions.

6.2 Unifying Themes

Although distinct in form and function, rods, rings, and shells reveal shared principles that unify the field of soft robotics. All three archetypes highlight the centrality of material compliance, nonlinear mechanics, and feedback-driven control. Each system demonstrates how bioinspired structures can achieve performance that rigid robots cannot—whether by exploiting fluid–structure coupling, orchestrating rhythmic actuation, or distributing pressure adaptively across a contact interface. Together, these archetypes provide a modular toolkit for designing robotic systems capable of safe interaction with complex environments.

6.3 Future Directions

The findings of this dissertation open several avenues for further exploration, spanning theory, fabrication, control, and application. While rods, rings, and shells have been treated as distinct archetypes, their convergence points to a broader design space where hybrid architectures and intelligent control strategies can redefine what soft robotic systems can achieve.

6.3.1 Advances in Modeling and Simulation

Future research should continue to refine the mathematical and computational models that underpin soft robotic design. While discrete elastic rod theory, continuum shell models, and fluid–structure interaction frameworks have proven effective, they remain computationally demanding for real-time implementation. Reduced-order models, physics-informed neural networks, and machine learning–augmented solvers may enable fast, accurate predictions of system dynamics while retaining critical nonlinear behaviors such as buckling, wrinkling, or peristaltic wave propagation. Furthermore, coupling these models with uncertainty quantification could allow soft robots to operate robustly in stochastic environments such as the human body or turbulent fluids.

6.3.2 Materials and Fabrication Innovations

The functionality of soft robots is intimately tied to advances in material science. Future directions include integrating bioresorbable polymers for temporary medical implants, hydrogels for enhanced biocompatibility, and electroactive polymers for untethered actuation. Multimaterial 3D printing and hybrid molding–casting techniques will allow precise control of spatial stiffness gradients, enabling devices that transition seamlessly from rigid to soft regions. Moreover, embedding stretchable electronics, energy storage, and wireless communication modules within rod-, ring-, and shell-like structures will allow for fully autonomous, minimally invasive robotic platforms.

6.3.3 Control and Autonomy

A key challenge remains the development of robust control frameworks capable of handling nonlinear, high-dimensional dynamics. Closed-loop controllers that fuse multi-modal sensing—force, strain, ultrasound, and vision—must evolve toward adaptive, learning-based strategies. Reinforcement learning, model predictive control, and bioinspired reflex loops offer promising paradigms for online adaptation in uncertain environments. Future work should also explore cooperative control among multiple soft robotic agents, enabling tasks

such as swarm-based navigation in complex vasculature or distributed manipulation in surgical contexts.

6.3.4 Biomedical Applications

Biomedical engineering continues to be a primary driver for innovation in soft robotics. Rod-like systems may evolve into microscale navigators for targeted drug delivery, minimally invasive biopsies, or precise catheter guidance in neurovascular procedures. Ring-based peristaltic actuators could be adapted for gastrointestinal diagnostics, robotic endoscopy, or assistive transport of delicate biological samples. Shell-inspired membranes offer tremendous promise for adaptive hemostasis, wound compression, and even dynamic organ support during surgery. Translational research should aim to bridge preclinical animal models with human trials, ensuring regulatory compliance, biocompatibility, and long-term safety.

6.3.5 Non-Medical Frontiers

Beyond the biomedical domain, future opportunities abound in environmental monitoring, exploration, and manufacturing. Rod-based robots may be deployed for subsurface inspection of confined pipelines or as untethered swimmers in search-and-rescue operations. Ring-based peristaltic systems could transport fluids or materials in space habitats, addressing waste management challenges in microgravity. Shell-based platforms may serve as conformal grippers for delicate agricultural harvesting, or as adaptive pads for collaborative industrial robots working safely alongside humans. These cross-domain applications underscore the versatility and scalability of the presented archetypes.

6.3.6 Toward Integrated Hybrid Systems

Perhaps the most exciting future direction lies in integration. Rods, rings, and shells should not be viewed as isolated archetypes but as complementary building blocks of multifunctional machines. A hybrid system might combine rod-like filaments for navigation, ring-like modules for transport, and shell-inspired membranes for conformal interaction. Such integration

could yield next-generation minimally invasive surgical robots capable of navigating tortuous vasculature, transporting therapeutic payloads, and applying adaptive compression—all within a single platform. Achieving this vision will require coordinated progress in modeling, materials, control, and clinical validation.

6.3.7 Ethical and Societal Considerations

As soft robotics progresses toward widespread biomedical and industrial deployment, it is important to consider ethical and societal implications. Patient safety, regulatory approval, environmental sustainability of materials, and equitable access to technology will play critical roles in shaping adoption. Collaborative frameworks involving engineers, clinicians, ethicists, and policymakers must be established to ensure that soft robotic technologies are deployed responsibly and inclusively.

In summary, the future of soft robotics is both technically challenging and rich with opportunity. The integration of advanced modeling, novel materials, intelligent control, and application-driven design holds the potential to transform how robots interact with biological and physical environments. By building on the foundational principles established in this dissertation, the field can move toward soft robotic systems that are not only adaptive and multifunctional, but also safe, accessible, and transformative across domains.

6.4 Closing Remarks

In summary, this dissertation has advanced the scientific understanding and practical design of soft robots inspired by rods, rings, and shells. By bridging biology, mechanics, and engineering, the work establishes a foundation for next-generation robotic systems that are not only flexible and adaptive but also clinically and scientifically impactful. The contributions underscore a central theme: that softness—long seen as a limitation in traditional robotics—can instead be a powerful enabler of safety, adaptability, and multifunctionality. The hope is that these principles will continue to guide the development of soft robotic tech-

nologies that transform both biomedical care and broader robotic applications in the years to come.

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