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# WHEN FRICTION MEETS FUNCTION

Engineering Medical Instruments  
with Friction to Glide, Grip and Propel



MOSTAFA A. ATALLA



# When Friction Meets Function

Engineering Medical Instruments with Friction to Glide, Grip and Propel

## Dissertation

for the purpose of obtaining the degree of doctor  
at Delft University of Technology  
by the authority of the Rector Magnificus, prof. dr. ir. H. Bijl,  
chair of the Board for Doctorates  
to be defended publicly on  
Tuesday, 1 September 2026, 10:00

by

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*to my family*





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# Summary

In medical instrument design, friction is traditionally treated as a binary choice: either minimize it to reduce wear and trauma, or maximize it to anchor and stabilize. Yet many transluminal tools, such as catheters and endoscopes, need both—low friction to navigate safely and high friction to remain stable at the target site. This dissertation reframes friction from a static constraint into a controllable parameter that can be actively modulated to engineer systems that glide, grip, and propel.

Part I of this dissertation investigates the use of ultrasonic lubrication to modulate friction at the instrument interface between low friction (glide) and high friction (grip) on demand. We use high-frequency transverse vibrations to pressurize the squeeze film of fluid at the contact interface, increasing the interfacial separation and lubricating the contact on demand. We develop a theoretical understanding of how this fluid-film pressure buildup occurs in liquid environments relevant to catheter applications, such as blood vessels, and use it to design catheter-compatible friction-control modules, demonstrating a friction reduction of up to 42% in *ex vivo* tests on porcine aorta and confirming the viability of ultrasonic lubrication for this application.

Part II of this dissertation shifts from local-level friction modulation to system-level tuning of friction anisotropy for robotic propulsion. We show that sequencing the motion of multiple sliding contact elements creates the system-level friction anisotropy required for self-propulsion, inspired by how parasitic wasps sequence the motion of their ovipositor valves (sliders) to penetrate skin, steer, and transport eggs with minimal external forces. We develop a theoretical framework linking motion patterns to thrust and stability and validate it through a multi-slider self-propelling robotic capsule for colonoscopy that uses a mechanically encoded slider-motion sequence for control-free propulsion in the colon.

Part III combines the local friction modulation introduced in Part I with the system-level tuning of friction anisotropy developed in Part II to enable efficient robotic locomotion. We demonstrate this combination in two bio-inspired locomotion systems based on inchworm locomotion and ovipositor-inspired transport. Both systems achieve locomotion using only two modules under identical normal loads and isotropic surface conditions, unlike traditional systems, which typically achieve locomotion using a larger number of modules or dedicated mechanisms for normal-load

modulation. These results underscore how coupling local and system-level friction tuning can improve locomotion efficiency, reduce design complexity, and unlock new possibilities for robotic propulsion.

By transforming friction from a passive constraint into an active control variable, this work shifts the design paradigm for medical instruments toward engineering with friction, not around it, enabling more optimal designs for physical interaction and unlocking a path toward a new class of friction-programmable instruments. Beyond medical applications, the methods developed in this dissertation provide a transferable toolkit for active friction control in soft robotics, human-robot interaction, and tribology, promising a future of mechanically intelligent, interaction-aware machines that adapt to the physical world with inherent safety and precision.

# Samenvatting

Bij het ontwerpen van medische instrumenten wordt wrijving traditioneel behandeld als een binaire keuze: óf je minimaliseert wrijving om slijtage en weefselschade te beperken, óf je maximaliseert wrijving om te verankeren en te stabiliseren. Toch hebben veel transluminale hulpmiddelen, zoals katheters en endoscopen, beide nodig—lage wrijving om veilig te navigeren en hoge wrijving om stabiel te blijven op de doellocatie. Dit proefschrift herpositioneert wrijving van een statische beperking naar een controleerbare parameter die actief kan worden aangepast om systemen te ontwerpen die kunnen glijden, grijpen en voortbewegen.

Deel I van dit proefschrift onderzocht het gebruik van ultrasone lubricatie om wrijving tussen de buitenzijde van het instrument en het omliggende weefsel op aanvraag aan te passen van laag (glijden) naar hoog (grijpen) en andersom. Door hoogfrequente transversale trillingen toe te passen om de dunne laag materiaal tussen de twee oppervlakten (i.e., squeeze-film) onder druk te brengen, genereerden we een onder druk staande vloeistoffilm die de scheiding tussen de twee oppervlakten vergroot en het contact op aanvraag smeert. We bouwden een theoretisch begrip op van hoe deze drukopbouw in de vloeistoffilm ontstaat in vloeibare omgevingen die relevant zijn voor kathetertoepassingen—zoals bloedvaten—en gebruikten dit om katheter-compatibele wrijvingscontrolemodules te ontwerpen. In *ex vivo*-testen op varkensaoorta demonstreerden we een wrijvingsvermindering tot 42%, waarmee we de haalbaarheid van ultrasone lubricatie voor deze toepassing bevestigden.

Deel II van dit proefschrift verschoof van lokale wrijvingsaanpassing naar het op systeemniveau afstemmen van anisotrope wrijving voor robotische voortstuwing. We lieten zien dat het afwisselend heen en weer bewegen van meerdere schuivende parallelle elementen de anisotrope wrijving op systeemniveau creëert die nodig is voor zelfvoortstuwing, geïnspireerd door hoe sluipwespen de beweging van de delen van hun legboor (sliders) afwisselend heen en weer te bewegen om met minimale externe krachten te penetreren, te sturen en eieren te transporteren. We ontwikkelden een theoretisch raamwerk dat bewegingspatronen koppelt aan stuwkracht en stabiliteit, en valideerden dit met een zelf voortstuwende robotcapsule met meerdere sliders voor colonoscopie die een mechanisch gecodeerde slider-bewegingssequentie gebruikt voor controle-vrije voortstuwing in de darmen.



Deel III van dit proefschrift combineerde lokale wrijvingsaanpassing (Deel I) met het op systeemniveau afstemmen van anisotrope wrijving (Deel II) om efficiënte robotische locomotie mogelijk te maken. We implementeerden deze combinatie in twee bio-geïnspireerde systemen voor locomotie gebaseerd op hoe wormen zich verplaatsen en hoe sluipwespen hun legboor voortbewegen en eieren transporteren, en realiseerden beweging met slechts twee modules onder identieke normaalkracht en isotrope oppervlaktecondities—een ongekend scenario in robotische locomotie. Dit onderstreept hoe het koppelen van wrijvingsafstemming op lokaal en systeemniveau de efficiëntie kan verhogen, de ontwerpcomplexiteit kan verminderen en nieuwe vormen van robotische locomotie mogelijk kan maken.

Door wrijving te transformeren van een passieve beperking naar een actieve regelvariabele, verplaatst dit werk het ontwerpparadigma voor medische instrumenten naar ontwerpen *met* wrijving in plaats van eromheen. Dit maakt optimalere ontwerpen voor fysieke interactie mogelijk en opent een pad naar een nieuwe klasse van instrumenten waarvan we de wrijving kunnen programmeren. Buiten medische toepassingen bieden de methoden die in dit proefschrift zijn ontwikkeld een overdraagbare gereedschapskist voor actieve wrijvingscontrole in zachte robotica, mens–robotinteractie en tribologie, met de belofte van een toekomst van mechanisch intelligente, interactiebewuste machines die zich met inherente veiligheid en precisie aanpassen aan de fysieke wereld.



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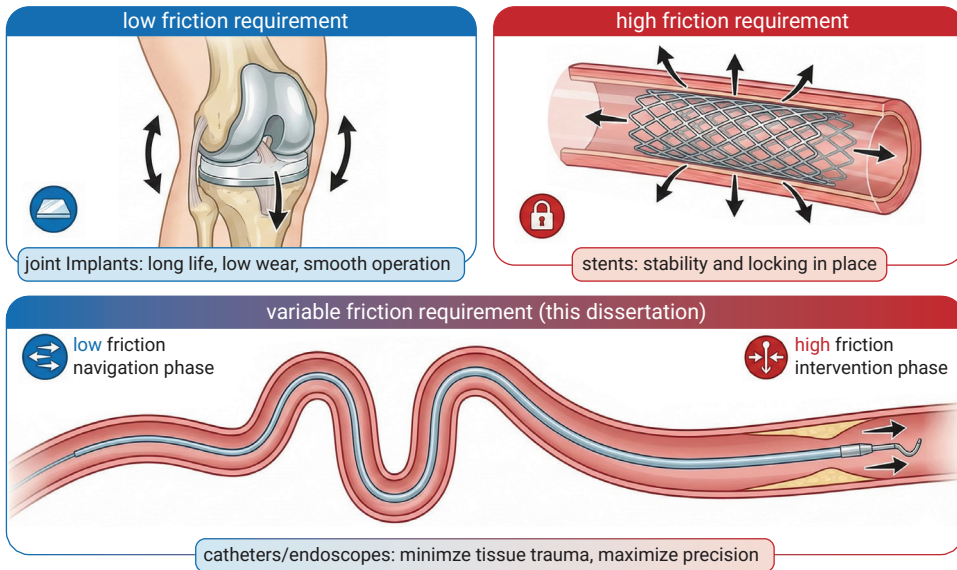
# Introduction

## 1.1. The Role of Friction in Medical Instruments

Medical instruments enable therapeutic interventions through mechanical interaction. Whether navigating a flexible catheter through the vasculature, placing a joint implant, or deploying a vascular stent, these instruments operate through physical contact, either with the surrounding tissue or across their own articulating surfaces. Friction at these contact interfaces is critical for the functionality of these devices, acting as both an enabling and a limiting factor: it provides the mechanical coupling needed to fulfill the clinical task, yet, if not controlled properly, it can compromise precision, or introduce excessive resistance leading to wear and tissue trauma.

To balance the enabling and limiting effects of friction, instruments must be carefully designed to meet the specific friction requirements of their clinical task. These requirements vary depending on the application and can be broadly used to classify instruments according to the friction level required at the interface. A first class of instruments requires low friction (Fig.1.1(a)): joint implants, for instance, demand ultra-low friction between their articulating surfaces to reduce wear, mimic the natural lubricity of cartilage, and prolong device lifetime. A second class of instruments requires high friction (Fig.1.1(b)): vascular stents use friction to anchor securely against the arterial wall and prevent migration. A third class of instruments requires both high and low friction within a single procedure (Fig.1.1(c)): endoluminal instruments, such as flexible catheters and endoscopes, are introduced through a small incision or natural orifice and navigated through anatomical pathways, such as blood vessels or the colon, until they reach the region of interest where the medical procedure is performed. During navigation, friction should be minimal to facilitate advancement and avoid tissue trauma. Conversely, once the target is reached and a clinical task such as stent deployment is performed, friction needs to be high enough to stabilize the instrument for accurate diagnosis and precise intervention.

While designing for a single friction requirement in the first two classes is relatively straightforward, engineering an endoluminal instrument to meet variable friction needs is challenging, as it requires the ability to switch between two opposing extremes of friction on demand. Current medical instrument design practices view friction as a fixed system property or a design constraint, largely set by the material and surface properties. Consequently, existing endoluminal instruments feature a fixed suboptimal friction level that is neither as low as desired during navigation for safer and smoother operation, nor as high as needed during intervention for maximum stability and precision. This suboptimal solution contributes to additional



**Figure 1.1. Friction requirements in medical devices.** Medical devices span a wide spectrum of frictional demands. Some devices require high friction to lock securely in place, others require low friction for smooth long-term operation, and a third class requires both low and high friction depending on the procedural stage. The top left panel shows devices that require persistently low friction, exemplified by joint implants, in which ultra-low interfacial friction minimizes wear, enables smooth articulation, and prolongs implant lifetime. The top right panel illustrates devices that rely on consistently high friction, such as vascular stents, where strong frictional anchoring between the stent struts and the vessel wall is essential to maintain positional stability and prevent migration. The bottom panel highlights instruments, such as catheters and endoscopes, with a variable friction requirement: during navigation, low friction with the wall is required to reduce insertion forces and tissue trauma, whereas during the intervention stage (for example, stent deployment), high friction with the wall is needed to stabilize the instrument, enhance positioning accuracy, and execution precision.

operational hurdles during navigation; as an instrument, such as a colonoscope, is pushed deeper into the tortuous anatomy of the colon, friction accumulates along its length, complicating instrument advancement and increasing the likelihood of instrument looping or buckling, which increase the risk of tissue damage and require advanced tools and skills to resolve.

As such, there is a critical need for mechanisms that can actively control friction to fulfill the contrasting friction requirements of endoluminal instruments on demand—switching between low friction (glide) for smooth navigation and high friction (grip) for stability, and to enable new functionalities, such as generating locomotion (propel) to help the instrument advance safely through torturous anatomy.

Addressing this need, this dissertation explores methods to turn friction of medical instruments from a passive design constraint into an active controllable variable (Fig.1.1(c)) that we can modulate on demand, to engineer systems (instruments) that can glide, grip, or propel.

## 1.2. The Physics of Friction

To understand how we can control friction, it is useful to begin with the classical macroscopic-scale mathematical description of sliding friction, known as the Coulomb-Amontons law of friction:

$$F_f = \mu N, \quad (1.1)$$

which relates the tangential friction force  $F_f$  experienced by an object when it starts to slide or during steady sliding against another surface, to the normal load  $N$  pressing the surfaces together through the proportionality constant  $\mu$ , the coefficient of friction—an effective scalar parameter that quantifies the apparent resistance to sliding for a given surface pair under specified conditions, as illustrated in Fig.1.2(a). This empirical law provides a macroscopic approximation of frictional interaction when contact conditions remain approximately unchanged in the relevant ranges of load, sliding speed, and time [1–5].

To connect the macroscopic description of friction, expressed by the Coulomb–Amontons law, to its physical origins, it is necessary to consider what happens at the interface on a microscopic scale. At that scale, contact occurs only in discrete micro-junctions, formed where the high points of surface roughness—known as asperities—come into contact, as shown in Fig.1.2(b). The sum of these discrete contact areas at these micro-junctions forms the true area of contact ( $A_{true}$ ), which is only a fraction of the apparent area of contact [1, 6, 7]. Accordingly, friction reflects the collective behavior of these micro-junctions as they form, sustain shear, and deform or rupture during sliding [1, 8].

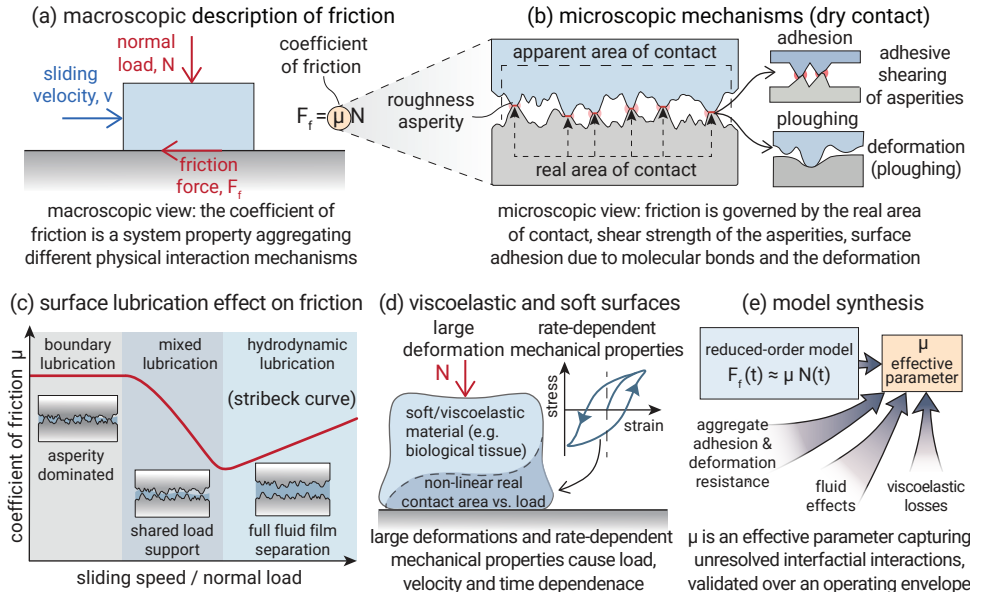
In dry contacts, friction is commonly attributed to two microscopic mechanisms: adhesion and deformation [1, 8, 9]. Adhesion arises when the mating asperities at micro-junctions stick due to intermolecular (van der Waals) forces, electrostatic attraction, or chemical bonding, depending on the materials and surface conditions [1, 10, 11]. This sticking results in an adhesive resistance force which must be overcome to initiate and sustain motion, approximated by  $F_{adhesive} = \tau \cdot A_{true}$ , where  $\tau$  is

the interfacial shear strength of the contact junctions—a measure of the amount of shear load the adhesive bonds can sustain before breaking [1, 8, 9]. The deformation mechanism arises from the mechanical interaction of these asperities. As they slide against one another, asperities undergo hysteretic deformation or irreversible plastic deformation, a process governed by surface hardness and material elasticity [1, 8, 9, 12]. This hysteretic or irreversible deformation dissipates energy which manifests itself as an additional resistance force  $F_{deform}$  [1, 8, 9, 13, 14].

Accordingly, the total friction force ( $F_f$ ) is the sum of the force required to shear the interfacial adhesive bonds ( $F_{adhesive}$ ) and the resistance arising from asperity deformation ( $F_{deform}$ ). The coefficient of friction ( $\mu$ ) is the total friction force normalized by the normal load  $\mu = F_f/N = (\tau \cdot A_{true} + F_{deform})/N$ . Despite the apparent dependence on  $N$ ,  $\mu$  is often approximately independent of normal load, because the friction force components ( $F_{adhesive} = \tau \cdot A_{true}$ ) and ( $F_{deform}$ ) tend to scale proportionally with ( $N$ ), keeping the ratio ( $F_f/N$ ) approximately constant. Consequently, the coefficient of friction ( $\mu$ ) depends mainly on the interfacial shear strength ( $\tau$ ), the true area of contact  $A_{true}$ , the surface hardness and the elasticity of the material [9, 12]. Accordingly,  $\mu$  is not a fundamental material property, but a system property driven by the interplay of these parameters [1, 5, 9, 14].

The presence of liquid at the interface introduces additional complexity. Unlike in dry contacts, a fluid film fundamentally alters the interaction: it supports the applied load by separating the surfaces, yet creates its own resistance as the fluid itself must be sheared during motion. At the initiation of sliding, the presence of liquid in the interfacial gaps can introduce capillary forces which can significantly increase the force required to initiate motion. As sliding initiates and speed increases, the thickness of the fluid film grows depending on the fluid viscosity ( $\eta$ ), sliding speed ( $v$ ), and normal load ( $N$ ) [15]. This change in film thickness determines the degree of separation between the two surfaces, which consequently affects their interaction and the resulting coefficient of friction ( $\mu$ ).

Depending on the film thickness during sliding, the coefficient of friction evolves through three different lubrication regimes as captured by the Stribeck curve shown in Fig.1.2(c) [16]. In the boundary lubrication regime (defined by a low speed to normal load ratio), surface interactions are dominated by asperity contact; here, the fluid is unable to separate the two surfaces and can only provide a molecular protective layer that reduces the interfacial shear strength ( $\tau$ ), thereby lowering the adhesive resistance ( $F_{adhesive}$ ). As the sliding speed increases, the fluid film begins to develop



**Figure 1.2. Overview of friction mechanisms and parameterization.** (a) Macroscopic description using the phenomenological Amontons–Coulomb law ( $F_f = \mu N$ ). (b) Microscopic view of dry contact, where real contact occurs at discrete asperities; dissipation is driven by adhesive shearing and deformation. (c) Fluid-mediated friction regimes (boundary, mixed, and hydrodynamic) illustrated by the Stribeck curve. (d) Mechanisms specific to soft and viscoelastic surfaces, where large deformations create nonlinear contact areas and bulk viscoelasticity introduces history-dependent hysteresis losses. (e) Modeling synthesis: Micro-scale adhesion, fluid effects, and bulk losses are aggregated into an effective friction parameter ( $\mu$ ) for reduced-order device analysis.

enough pressure to partially separate the surfaces, known as the mixed lubrication regime. In this regime, the fluid film carries a portion of the normal load, which decreases the number of asperities in contact, consequently reducing the true area of contact ( $A_{true}$ ) and the coefficient of friction. Eventually, the system reaches the hydrodynamic lubrication regime, characterized by full fluid-film separation where the surfaces are no longer in physical contact and friction arises solely from the viscosity of the fluid [17, 18].

Biological interfaces introduce an additional layer of complexity that deviates from the idealized friction mechanisms introduced earlier. Unlike rigid surfaces, biological tissues are typically soft and viscoelastic, meaning they deform like elastic solids but also dissipate energy like viscous fluids. These characteristics fundamentally alter the friction understanding established for rigid bodies: the high compliance of the tissue allows for significant surface deformation even under light loads, causing the true area of contact ( $A_{true}$ ) to increase nonlinearly. Furthermore, viscoelasticity intro-

duces rate-dependent hysteresis losses, where energy is dissipated through internal deformation and recovery of the bulk material itself rather than solely at the interface [19–21]. Consequently, the resulting coefficient of friction ( $\mu$ ), when an instrument is in contact with tissue, can exhibit a complex dependence on load, velocity, and time [4, 22].

However, this added complexity in biological interfaces does not preclude the use of the simplified Coulomb-Amontons law as a framework to model frictional interaction with tissue. In this context, the law serves as a reduced-order model, a deliberately simplified description capturing dominant effects, when validated over the relevant conditions [1, 2, 4]. Within this framework,  $\mu$  is treated as an effective parameter that represents unresolved energy dissipation mechanisms, including the nonlinear contact and viscoelastic hysteresis characteristics of biological surfaces, provided they remain approximately consistent over the operating envelope of interest [1, 19, 20]. Friction control then becomes the problem of regulating the factors that set friction: the normal load and the effective  $\mu$  (via interfacial shear strength, lubrication state, and true contact area), forming the basis for the active friction control strategies developed in this dissertation.

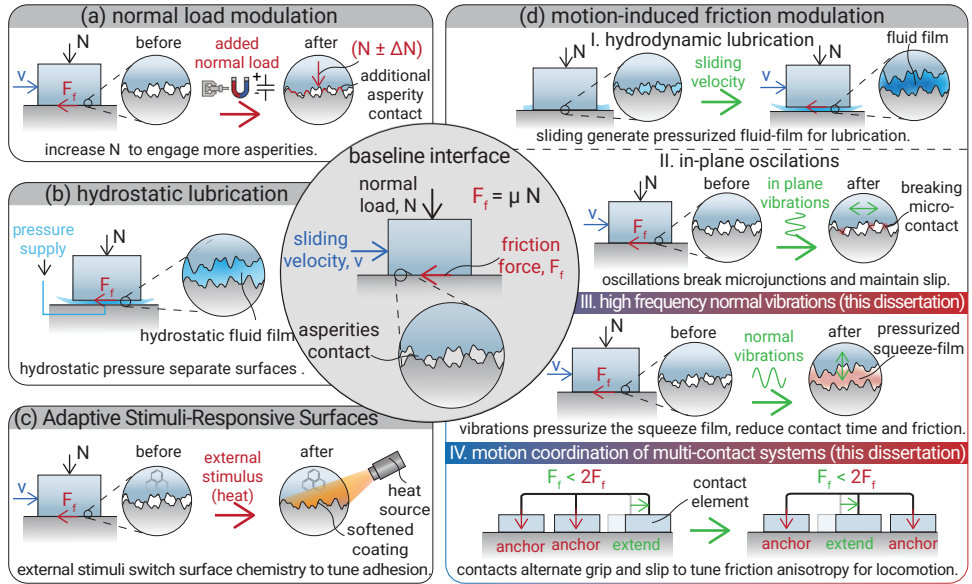
### 1.3. Active Friction Control Strategies

Active friction control strategies can be classified based on the control parameter. At a high level, these strategies can be divided into two main methods: those that modulate the normal load ( $N$ ) and those that modulate the coefficient of friction ( $\mu$ ). As established in the previous section,  $\mu$  encapsulates multiple physical effects: interfacial shear strength  $\tau$ , true area of contact  $A_{true}$ , and lubrication condition, each of which presents a potential control knob of friction. In the following subsections, we examine these strategies in turn, beginning with methods for Normal Load Modulation, followed by strategies to control the physical parameters associated with  $\mu$ , which we classify based on their actuation mechanism: Hydrostatic Lubrication, Adaptive Stimuli-Responsive Surfaces, and Motion-Induced Friction Modulation .

#### 1.3.1. Normal Load Modulation

Normal load modulation strategies rely on the direct scaling relationship between the friction force ( $F_f$ ) and the applied normal load ( $N$ ), as described by the Coulomb-Amontons law ( $F_f = \mu N$ ). This strategy (Fig. 1.3(a)) is commonly implemented





**Figure 1.3. Overview of active friction control strategies** The central panel illustrates the baseline interface governed by Coulomb's law ( $F_f = \mu N$ ), where resistance arises from direct asperity contact. Active strategies manipulate either the normal load or the coefficient of friction to control the friction force: (a) Normal load modulation directly alters the applied force ( $N$ ) to change the degree of asperity engagement. (b) Hydrostatic lubrication utilizes external supply pressure to create a load-bearing fluid film that separates the sliding surfaces. (c) Adaptive stimuli-responsive surfaces use external triggers (e.g., heat) to tune intrinsic interfacial properties, such as surface adhesion or coating stiffness. (d) Motion-induced strategies leverage relative velocity or vibrational dynamics to modify the effective contact state, including (I) hydrodynamic fluid film lubrication via relative sliding, (II) slipping via in-plane oscillations, (III) squeeze-film lubrication via high frequency normal vibrations, and (IV) system-level friction anisotropy tuning via motion coordination. The strategies highlighted in red and blue (III and IV) are the methods investigated in this dissertation.

through non-contact field modulation, such as electroadhesion or magnetic actuation, where an external electrostatic or electromagnetic source generates a controllable attraction force across the interface [23, 24]. Alternatively, mechanical actuation strategies employ pneumatic, hydraulic, or vacuum-based systems to modulate the normal force, mimicking biological mechanisms such as the reflexive grip adjustment in humans to prevent objects from slipping or the radial expansion of earthworms to press against tunnel walls forming frictional anchors for locomotion [25].

While field-based methods offer high bandwidth and rapid reversibility, their practical application is often limited by achievable force density and sensitivity to environmental conditions such as humidity or surface roughness. Conversely, mechanical approaches can generate large normal forces, however; they typically require additional mechanisms and actuation, increasing the overall complexity of the system.

Consequently, normal load modulation is often a trade-off between the required dynamic range of friction and the physical constraints of the actuation hardware.

### 1.3.2. Hydrostatic Lubrication

Hydrostatic lubrication strategies alter the effective coefficient of friction ( $\mu$ ) by forcing a lubricant into the contact interface using an external pressure source, creating a load-bearing film that separates the surfaces even when the system is stationary, as shown in Fig.1.3(b). This separation is typically achieved through micro-channels or porous surface layers that distribute fluid under a controlled supply pressure ( $P_{supply}$ ), effectively governing the film thickness ( $h$ ) and preventing direct solid-to-solid contact [18]. As such, the system can maintain a low-friction state regardless of motion, compensating for leakage by actively regulating the flow rate. Although this approach guarantees ultra-low friction even under heavy normal loads, it requires bulky hydraulic infrastructure (pumps and reservoirs), limiting its integration into compact or mobile devices.

### 1.3.3. Adaptive Stimuli-Responsive Surfaces

Adaptive stimuli-responsive strategies modulate friction by altering the physicochemical properties of the interface itself to control the adhesive resistance ( $F_{adhesive} = \tau \cdot A_{true}$ ) [26], as shown in Fig.1.3(c). This modulation is typically achieved through stimuli-responsive coatings, such as polymer or hydrogel coatings, that undergo reversible structural changes at the interfacial level in response to external triggers like electric fields, temperature, or pH [27, 28]. For instance, a collapsed polymer chain may present a high-friction, adhesive surface, whereas a swollen, hydrated state can drastically reduce  $\tau$ . Similarly, surfaces with switchable wettability can dynamically alter the capillary forces between asperities, turning the "stickiness" of the interface on or off by modifying the contact angle of the interfacial fluid [29, 30].

While utilizing stimuli-responsive coatings allows for precise, localized control of friction across a surface without adding physical bulk to the instrument, this chemical approach suffers from severe practical limitations in physiological environments. Because this modulation relies on delicate surface chemistry, the magnitude of the friction change is highly sensitive to surrounding conditions and can be easily disrupted by biological fluids or contamination. Furthermore, the effectiveness of this approach

is fundamentally restricted by the slow response time of the chemical transitions and the strict biocompatibility requirements of the materials.

### 1.3.4. Motion-Induced Friction Modulation

Motion-induced strategies modulate the effective coefficient of friction ( $\mu_{eff}$ ) by using relative motion at the interface to alter the contact mechanics (Fig. 1.3(d)). A common mechanism in this category is hydrodynamic lubrication which uses relative sliding velocity to generate a fluid film pressure that supports a portion of the normal load ( $N$ ) and reduces asperities contact [18, 31]. Alternatively, high-frequency vibrations, typically in the ultrasonic range, can actively manipulate the contact state: oscillations perpendicular to the contact surface generate squeeze-film pressure to separate surfaces or induce intermittent contact, effectively reducing the time-averaged true area of contact ( $A_{true}$ ) [32–34]. Similarly, rapid in-plane oscillations continuously disrupt the formation of adhesive junctions, maintaining a low-resistance slip condition [35].

The principle of motion-induced friction modulation extends to the macroscopic system level, where multi-contact systems regulate the system level friction interaction with the environment through the coordinated anchoring, releasing, and extending of the contact elements [36, 37]. Consequently, the effective system-level friction is determined by the motion pattern and time sequencing of the contact elements, allowing for dynamic tuning of friction anisotropy.

## 1.4. Friction Management in endoluminal Medical Instruments

Current friction management in transluminal interventions relies almost exclusively on passive, uncontrollable strategies. To facilitate navigation and minimize tissue trauma, conventional devices utilize liquid lubricants [38] or self-lubricating hydrophilic coatings that become lubricious upon exposure to physiological fluids [39]. However, these methods suffer from notable physical limitations. Add-on liquid lubricants are inherently short-lived, gradually wiping away and losing effectiveness as the instrument advances. Hydrophilic coatings are effective in specific fields such as urology, however, they lack the versatility required for broader endoluminal applications. Crucially, neither approach offers dynamic control; they cannot transition the instrument between low and high friction states based on the specific demands

of the procedural phase, further highlighting the need for an active friction control strategy in endoluminal instruments.

## 1.5. Dissertation Aims

This dissertation aims to overcome the fundamental limitations of fixed frictional properties in current endoluminal medical instruments. Relying on these fixed properties results in a suboptimal interaction between the instrument and the surrounding tissue. To address this limitation, this dissertation investigates motion-induced friction modulation methods to achieve active friction control in endoluminal instruments. This approach enables dynamic friction control that is largely independent of the interfacial material properties or surface design, potentially allowing for seamless integration with existing instruments.

The dissertation investigates two specific, complementary motion-induced friction modulation methodologies: high frequency transverse vibrations and motion coordination of multi-contact systems. High frequency transverse vibrations enable direct control over the coefficient of friction at the contact interface, which can potentially address the variable friction needs of endoluminal instruments, allowing them to switch on demand between low friction for smooth navigation (glide) and high friction for stable intervention (grip). Motion coordination of multi-contact systems enables system-level active tuning of frictional anisotropy, which can potentially address the need for a control method to actively assist navigating through torturous anatomy (propel). Guided by these selected methods, this dissertation addresses the following specific aims:

1. **Investigate the viability of high-frequency transverse vibrations for friction modulation in liquid-submerged environments and its translation into a functional endoluminal instrument design.** This aim investigates whether the fundamental physical mechanisms of vibration-induced friction reduction remain effective under the wet physiological conditions required in clinical applications, specifically for catheters operating within blood vessels, and explores the translation of these principles into a physical device architecture to evaluate the effectiveness of the technique when in contact with biological tissue interfaces.
2. **To explore system-level friction anisotropy tuning methods and characterize bio-inspired motion coordination patterns for endoluminal locomotion.** This aim explores bio-inspired methods for friction anisotropy tuning through

motion coordination, and investigates the relationship between the motion pattern and the resulting frictional anisotropy to understand how these patterns can be leveraged to generate active self-propelling through tortuous anatomy.

3. **To explore the coupling of local-level vibratory modulation with system-level motion coordination within a unified architecture.** This aim explores the feasibility and potential benefits of combining both local-level friction control and system-level friction anisotropy tuning. The objective is to determine whether a single integrated design can successfully consolidate the targeted functionalities—glide, grip, and propel—and investigate its benefits for robotic locomotion.

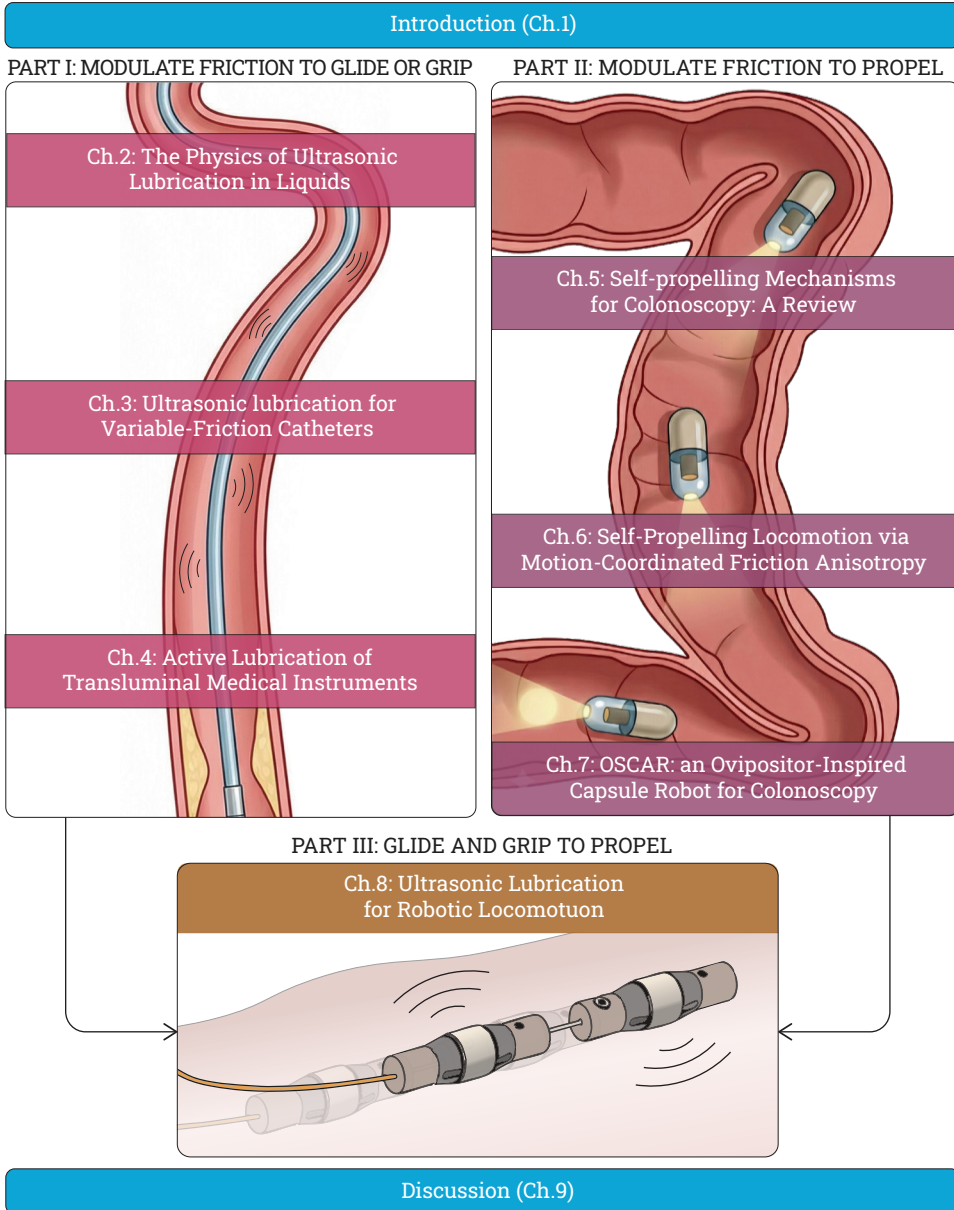
## 1.6. Dissertation Layout

The dissertation is structured in three parts which reflect the dissertation aims, as shown in the dissertation layout in Fig. 1.4:

**Part I: Modulate Friction to Glide or Grip** (Chs. 2–4) develops ultrasonic-lubrication-based variable-friction tools. Chapter 2 formulates the physical basis using incompressible squeeze-film levitation. Chapter 3 builds toward variable-friction catheters using ultrasonic lubrication, with emphasis on how vibration can reduce friction on demand in flexible tools. Chapter 4 extends these concepts to active lubrication of intraluminal medical instruments and discusses practical integration considerations.

**Part II: Modulate Friction to Propel** (Chs. 5–7) develops friction-enabled self-propelling devices. Chapter 5 reviews self-propelling robots for colonoscopy and frames the design requirements for propulsion in compliant lumens. Chapter 6 introduces a mechanically inflatable, bio-inspired locomotion concept for robotic pipeline inspection and evaluates how it generates motion through frictional interactions. Chapter 7 presents OSCAR, an ovipositor-inspired capsule robot for colonoscopy, and demonstrates how friction anisotropy and coordinated motion of a multi-contact self-propelling system can tune the system-level frictional anisotropy to achieve locomotion in confined anatomy.

**Part III: Glide and Grip to Propel** (Ch. 8) combines local-level friction control (Part I) with system-level friction anisotropy tuning (Part II) to achieve locomotion. (Ch. 8)



**Figure 1.4. Dissertation structure and layout.** Following the introduction, the dissertation is organized into three parts corresponding to the specific research aims. **Part I** (Chapters 2–4) addresses the first aim by investigating local-level friction modulation (ultrasonic vibrations) to achieve *glide or grip*. **Part II** (Chapters 5–7) addresses the second aim by exploring system-level friction modulation (multi-contact motion coordination) to *propel*. **Part III** (Chapter 8) addresses the third aim by integrating these methodologies, coupling the *glide and grip* mechanisms of Part I with the locomotion strategies of Part II to demonstrate a unified architecture, followed by a concluding discussion.

introduces two bio-inspired locomotion use cases that utilize the combined friction control approach to validate the concept and assess its benefits.

The final discussion chapter (Ch. 9) synthesizes the insights from the three parts and outlines the broader implications of the findings of this dissertation on the future of medical instruments design and beyond.

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# PART I

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# Modulate Friction to Glide or Grip

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# CHAPTER 2

## The Physics of Ultrasonic Lubrication in Liquids

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## Abstract

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When surfaces in contact undergo transverse ultrasonic vibration, the trapped fluid at their interface is nonlinearly compressed, producing a pressurized “squeeze film” that pushes the surfaces apart—a phenomenon known as squeeze-film levitation. By creating a fluid-lubricated interface that minimizes solid-to-solid contact, this levitation effect significantly reduces friction during relative sliding, forming the physical basis of ultrasonic lubrication. Because endoluminal instruments, such as catheters, operate in the presence of bodily fluids, for example blood in vascular applications, understanding how this levitation effect behaves in incompressible liquids is essential to explore if and how ultrasonic lubrication can be used to actively control friction in endoluminal medical applications. In this chapter, we investigate the underlying physics of squeeze-film levitation in incompressible fluids. Whereas in classical applications operating in dry environments, transverse vibrations are known to induce levitation through the nonlinear compressibility of air, classical Reynolds lubrication theory predicts no steady-state levitation in incompressible liquids, contrary to prior experimental observations. To resolve this discrepancy, we relax the classical assumption that fluid inertia is negligible and demonstrate that fluid inertial effects are the primary origin of the load-carrying capacity of incompressible squeeze films required for levitation. On this basis, we derive the physical law governing the steady-state levitation height as a function of vibration amplitude and frequency, contact area, applied force, and fluid density, providing the theoretical foundation for the variable-friction catheter concepts developed in the subsequent chapters.

## 2.1. Introduction

When an object is placed in closed proximity to a surface vibrating at sufficiently high frequency, it levitates. Levitation has been exploited in a wide range of applications from squeeze-film bearings [2, 3], contactless manipulation and transportation of objects [4, 5] to friction-modulation in surface haptics touchscreens [6, 7].

In all those examples, levitation occurs when the surrounding fluid is air, which is compressible, via the so-called squeeze-film effect. Having the same ability to levitate objects in incompressible liquid environments could help bring non-contact manipulation and friction modulation to a range of applications, including the medical field where blood and liquids are omnipresent.

In-liquid levitation has been addressed by a limited number of studies. Hatanaka et al. [8] experimentally demonstrated that underwater near-field levitation is possible. Nomura et al. [9] realized a noncontact transportation underwater using ultrasonic traveling waves. However, neither of these studies provided an explanation of the physical underpinnings of this phenomenon, with the second speculating that it might be due to the non-linear viscosity of liquid. Tamura et al. [10] hypothesized that underwater levitation is due to formation of cavitation bubbles, which collapse at the surface of the levitated object. Although their experimental data matches well with their hypothesis, the relatively large experimental film thickness ( $> 100 \mu\text{m}$ ) and high power ultrasonic transducer (350 W) limits the generality of their conclusion. This evidence indicates a gap in understanding the physical principles behind in-liquid levitation and thus, requires revisiting the existing theories.

The first leading theory that models squeeze film levitation is the Reynolds' lubrication theory [11]. This theory approaches the problem from the perspective of viscous fluids in a flow regime with negligible fluid inertia. Langlois [12] mathematically formalized the theory for the isothermal squeeze film case, which was later adopted and validated experimentally by Salbu [13] who showed that air squeeze films levitates mainly due to the non-linear compressibility of the viscous air film. Clearly, such an explanation is not applicable anymore once a liquid is used. As a consequence, employing the Reynolds equation to model the in-liquid levitation phenomenon, fails and predicts no steady state levitation force. Based on this result, Stolarski et al. [14] believed that oil squeeze films have no load-carrying capacity. However, this theoretical result clearly contradicts the experimental evidence, which disqualifies the lubrication theory in its current form from modeling liquid squeeze film levitation.



The second leading theory in modeling squeeze film levitation is the acoustic radiation pressure theory. This theory takes the perspective of wave propagation in compressible inviscid fluids. The pioneering work of Chu and Apfel [15] shed the light on the radiation pressure of compression waves propagating through an elastic fluid medium in a confined space, acting on a perfectly reflecting surface. This fundamental work was adopted and simplified later by Hashmimoto et al. [16] to model the special case of air thin films. Zhao et al. [17] showed experimentally, however, that this theory fails to capture the physics in air thin films of a typical thickness ( $< 100 \mu\text{m}$ ), which was further confirmed by the experimental results of Li et al. [18]. This is mainly due to the fact that the viscous penetration depth (i.e. boundary layer thickness) is in the same order of magnitude of the film thickness, suggesting significant viscous effects which cannot be neglected [19]. The viscous effects of the boundary layer become even more significant in liquid whose viscosity is around two orders of magnitude higher than its air counterpart, which rules out the applicability of this theory to model the behavior of liquid squeeze films.

A limited number of studies attempted to derive a unified viscoacoustic theory that works across the viscous and acoustic regimes. Melikhov et al. [20] developed a viscoacoustic model and identified the different operating regimes for air squeeze films as a function of the levitation height, confirming a purely viscous regime for typical squeeze film levitation systems. Ramanarayanan et al. [21] proposed another unified theory which described critical parametric conditions that causes levitation forces to switch to adhesion forces in air squeeze film systems. Remarkably, in the incompressible limit, their formulation predicted only weak adhesive squeeze film forces, adding even more uncertainty around the behavior of incompressible squeeze films.

In other terms, the existing theories fail to capture the physics of liquid squeeze film levitation. We can also conclude that a viscous fluid approach (i.e. similar to the lubrication theory) is essential to tackle this problem given the comparable size of the film thickness and the boundary layer. However, relaxation of the assumptions of the lubrication theory is needed to uncover the underlying physics behind this phenomenon and find out its physical governing law.

In this chapter, we show that a stable steady-state levitation of objects can be obtained in incompressible liquid environments at ultrasonic vibration frequencies. We uncover the fundamental pressure-inducing mechanisms in incompressible films. We reveal the physical law that governs the steady-state levitation height, which

we validate experimentally. Finally, we compare the load-carrying capacity of liquid squeeze films with its air counterpart.

## 2.2. Modeling

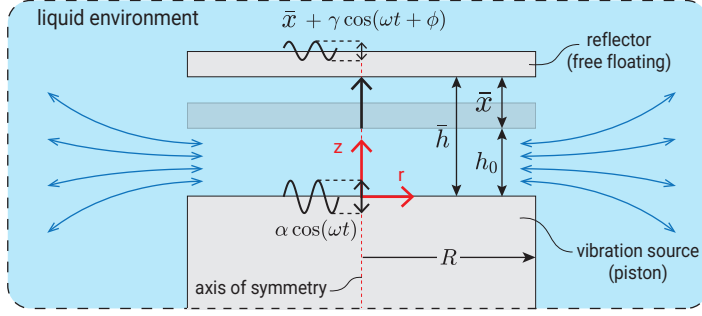
Consider an axisymmetric system of a sinusoidal vibration source, a free-floating reflector and a thin film of liquid in between as shown in Fig.2.1. We define the time-dependent film thickness  $h(t)$  such that  $h(0) = h_0$ , where  $h_0$  is an assumed initial film thickness between the source and reflector. The disc radius is  $R$  such that  $R \gg h(t)$ . The source oscillates with an angular frequency  $\omega$  and an amplitude  $\alpha$  such that  $\alpha < h(t)$ . We define the squeeze Reynolds number [22] to be  $Re_s = \rho \omega \bar{h}^2 / \mu$  where  $\bar{h}$  is the steady-state time-averaged film thickness,  $\rho$  is the density of the liquid and  $\mu$  is its viscosity. This number gives a measure of the relative significance of inertial and viscous effects.

One of the key assumptions of the lubrication theory is that fluid inertia is negligible and thus fluid behavior is dominated by viscous effects. This proved to be true for gas squeeze films since the squeeze Reynolds number is typically less than unity ( $< 1$ ) [23, 24]. However, in the case of a typical liquid, such as water, the order of magnitude of the physical parameters of the Reynolds number is as follows:  $\rho \approx O(10^3)$  Kg/m<sup>3</sup>,  $\omega \approx O(10^4)$  rad/s and  $\mu \approx O(10^{-3})$  Pa.s. Given the experimental data of Hatanaka et al. [8], we can also expect the order of magnitude of  $(\bar{h})$  to be  $\bar{h} \approx O(10^{-5})$  m. This yields a Reynolds number of order  $Re_s \approx O(10^1)$  suggesting the significance of the film inertia and thus, the lubrication theory's assumption becomes invalid. Therefore, we revisit the basic fluid governing equations, which take inertial effects into account.

Given the aforementioned axisymmetric system of a source and reflector and given that  $R \gg h(t)$ , we can safely assume that the pressure gradient across the film thickness is negligible compared to the radial pressure gradient ( $\frac{\partial p}{\partial z} \ll \frac{\partial p}{\partial r}$ ) and therefore the pressure is only a function of the radial coordinate ( $\frac{\partial p}{\partial z} = 0$ ). With this assumption, the conservation of momentum and mass equations for this system using cylindrical coordinates, are given by [25]:

$$\rho \left( \frac{\partial v_r}{\partial t} + v_r \frac{\partial v_r}{\partial r} + v_z \frac{\partial v_r}{\partial z} \right) = -\frac{\partial p}{\partial r} + \mu \frac{\partial^2 v_r}{\partial z^2} \quad (2.1)$$

$$\frac{1}{r} \frac{\partial}{\partial r} (r v_r) + \frac{\partial v_z}{\partial z} = 0 \quad (2.2)$$



**Figure 2.1. In-liquid squeeze film levitation.** Initially, a film thickness  $h_0$  separates the vibrator and the free-floating reflector. This initial film thickness originates from various sources such as the roughness of the two surfaces or misalignment. Upon the start of the vibration  $\alpha \cos(\omega t)$ , a pressure builds up in the film layer, which pushes reflector away distance  $x$  until it reaches equilibrium at the time-averaged steady-state levitation distance  $\bar{x}$ . The reflector oscillates around its equilibrium position harmonically, denoted by  $\gamma \cos(\omega t + \phi)$ . The over-pressure in the liquid layer is linked to the steady state time-averaged levitation film height.

where  $p$  is the liquid pressure,  $\rho$  is its density,  $\mu$  is its viscosity, and  $v_r$  and  $v_z$  are the liquid velocity fields along  $r$  and  $z$  respectively. To solve for the pressure and velocity fields, we follow the iterative scheme of Kuzma and Jackson [26, 27]; we first approximate the inertial forces (left hand side of Eq.2.1) by employing the velocity profiles of the classical lubrication theory, which are given by:

$$v_r = \frac{3r\dot{h}}{h^3} (z^2 - hz) \quad (3a)$$

$$v_z = -\frac{\dot{h}}{h^3} (2z^3 - 3hz^3) \quad (3b)$$

Then, by integrating Eq.2.1 twice with respect to  $(z)$  and assuming static boundary conditions ( $v_r(r, 0, t) = v_r(r, h, t) = 0$ ), we find the following new expression for the radial velocity field ( $v_r$ ):

$$v_r = \frac{1}{\mu} \frac{\partial p}{\partial r} \left( \frac{z^2}{2} - \frac{hz}{2} \right) + \frac{\rho}{\mu} \left[ \frac{3r\ddot{h}}{h^3} \left( \frac{z^4}{12} - \frac{hz^3}{6} + \frac{h^3z}{12} \right) + \frac{r\dot{h}^2}{h^6} \left( -\frac{z^6}{10} + \frac{3hz^5}{10} - \frac{3h^2z^4}{4} + h^3z^3 - \frac{9h^5z}{20} \right) \right] \quad (4)$$

The conservation of mass principle requires that the inflow or outflow across the control volume of the film is equal to the volume change due to the source vibration and the reflector levitation. This condition can be expressed mathematically in integral form as follows:

$$\int_0^h v_r dz = -\frac{r\dot{h}}{2} \quad (5)$$

Finally, by substituting the velocity profile expression Eq.4 into the continuity equation Eq.5, we obtain the following expression for the pressure gradient along  $(r)$  [26]:

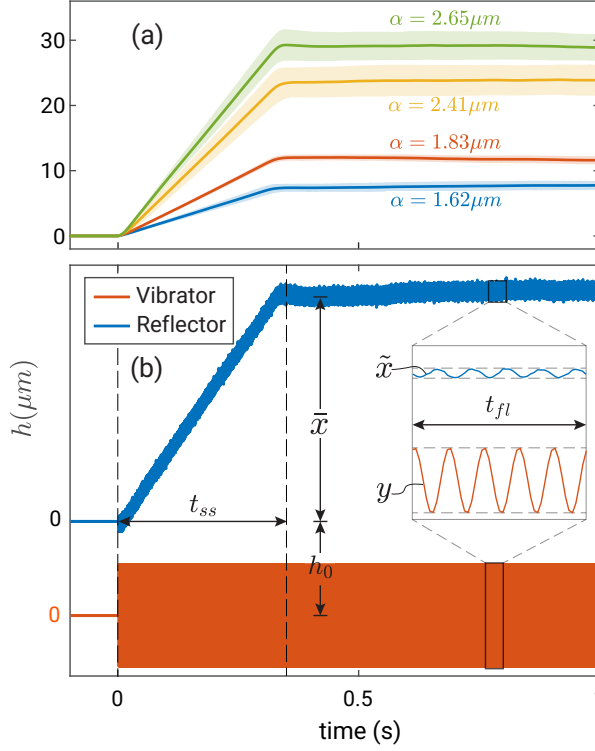
$$\frac{\partial p}{\partial r} = \frac{6\mu r\dot{h}}{h^3} + \frac{3\rho r\ddot{h}}{5h} - \frac{15\rho r\dot{h}^2}{14h^2} \quad (6)$$

This expression of the pressure gradient has different coefficients of the second and third terms compared to the one presented by Li et al [28], because in that work the term  $(v_z \frac{\partial v_r}{\partial z})$  was neglected. However, the magnitude of this term was previously shown to be comparable to  $(v_r \frac{\partial v_r}{\partial r})$  [26]. By substituting Eq.6 back into Eq.4, we can obtain an improved expression for  $(v_r)$  which, in turn, can be substituted in Eq.2.2 to obtain a new expression for  $(v_z)$ . This iterative process can be repeated by substituting the new expressions of  $(v_r)$  and  $(v_z)$  into the left hand side of the momentum equation Eq.2.1 and then move through the same steps. However, Grimm et al. [29] previously concluded that this first order perturbation solution is stable and showed close agreement with the full numerical solution unlike the higher order perturbation solutions [30]. Thus, repeating the iterative scheme is of no benefit here. By integrating the pressure gradient expression with respect to  $(r)$  assuming the pressure boundary condition  $(p(R, t) = p_a - \Delta p)$  where  $p_a$  is the ambient pressure and  $\Delta p$  is a pressure loss term due to the edge effect (to be discussed later), we obtain the following pressure field expression:

$$p - p_a = \frac{r^2 - R^2}{2} \left( \frac{6\mu\dot{h}}{h^3} + \frac{3\rho\ddot{h}}{5h} - \frac{15\rho\dot{h}^2}{14h^2} \right) - \Delta p \quad (7)$$

$$\text{where } \Delta p = \begin{cases} \frac{C_e \rho R^2 \dot{h}^2}{8h^2} & \text{if } \dot{h} > 0 \\ 0 & \text{otherwise} \end{cases}$$

This pressure profile expression suggests three main pressure-inducing mechanisms for incompressible liquid films. The first one is associated with the viscosity



**Figure 2.2. Experimental measurements of in-liquid squeeze-film levitation.** (a) demonstration of the stable steady-state levitation of a free floating reflector of mass 64g, that we can achieve under sinusoidal excitation of frequency 40kHz at different vibration amplitudes. For each vibration amplitude, five trials of measurements were collected and the results are plotted as the mean displacement (solid lines) and the standard deviation (shades). (b) A detailed view of one experimental trial corresponding to the input vibration amplitude  $\alpha = 1.62 \mu\text{m}$ , with the vibrator's displacement being  $y = \alpha \cos(\omega t)$ . We observe that the floating mass levitates to a steady-state position  $\bar{x}$  and oscillates around the equilibrium position in response to the vibration input, where the oscillations are represented as  $\tilde{x}$ . We observe that the amplitude of the reflector oscillations is in the same order of magnitude as the input amplitude. The response also shows two characteristic time scales: (a) time corresponding to the mass reaching steady state levitation distance  $t_{ss}$  (b) time associated with the reflector oscillations around its equilibrium position  $t_{fl}$ .

of the liquid represented by the term  $6\mu\dot{h}/h^3$ . This term is symmetric and thus, has no contribution to the time-averaged pressure. The second and third mechanisms are associated with the temporal and convective accelerations of the liquid squeezing in and out of the thin film, represented by the asymmetric terms  $3\rho\ddot{h}/5h$  and  $15\rho\dot{h}^2/14h^2$  respectively. In addition, accounting for liquid inertia results in an inevitable pressure drop  $\Delta p$  at the edge of the squeeze film during negative squeeze motion ( $\dot{h} > 0$ ), due to the sudden contraction of the liquid at the film entrance. Using Bernoulli's equation, we can find an expression for the pressure drop term [31, 32]

equal to  $C_e \rho R^2 \dot{h}^2 / 8h^2$  where  $C_e$  is a pressure loss coefficient that depends mainly on the geometry of the film entrance (i.e. sharp or round edge). This edge effect term along with the asymmetric temporal and convective acceleration terms contribute to a non-zero time-averaged pressure in the liquid film, thus clarifying that the in-liquid levitation phenomenon is driven by the inertial forces of the liquid.

By looking closely at the pressure profile expression, we notice that the film pressure depends mainly on the time-varying film thickness  $h(t)$  and its time derivatives  $\dot{h}, \ddot{h}$ . In a free-floating reflector context, the film thickness variables will depend on the relative motion between the source and reflector. Therefore, it is essential to couple the fluid dynamics with the dynamics of the levitated mass. The importance of dynamics coupling was also noted previously by Minikes et al. [33] who reported that the dynamics of the floating mass has significant influence on the behavior of the fluid film. Incorporating the dynamics is also beneficial since for almost all applications (e.g. squeeze film bearings), the main interest is to levitate objects freely, not to fix their positions. Thus, predicting the levitation height directly is of paramount importance from an application perspective.

To incorporate the levitated object dynamics, we assume a single degree of freedom system where the reflector is a mass. The mass is connected to the vibrating surface through a liquid layer whose initial thickness is  $h_0$ . We introduce two independent coordinates  $(x, y)$  where  $x$  is the displacement of the levitated object and  $y$  is the displacement of the vibrating surface. The film height  $h(t)$  is a function of  $x$  and  $y$  coordinates as follows:

$$h = h_0 + x - y, \quad \dot{h} = \dot{x} - \dot{y}, \quad \ddot{h} = \ddot{x} - \ddot{y} \quad (8)$$

By analyzing the forces acting on the reflector, we can obtain its equation of motion and arrange the terms to get the following expression for its acceleration:

$$\ddot{x} = \left( \frac{10\dot{h}^2}{14h^2} - \frac{2\ddot{y}}{5h} - \frac{4\nu\dot{h}}{h^3} - \frac{C_e\dot{h}^2}{4h^2} - Mg \right) / \left( M + \frac{2}{5h} \right) \quad (9)$$

where  $M = \frac{m}{\pi^2 R^4 \rho}$ ,  $\nu = \frac{\mu}{\rho}$

where the mass ratio ( $M$ ) is the ratio of the mass per unit area of the levitated object, to the mass per unit height of the liquid film. It is a measure of the relative significance of the inertial effects of the free floating object and the liquid film. The kinematic

viscosity ( $\nu$ ), on the other hand, is a measure of the fluid resistance to flow under inertial forces.

## 2

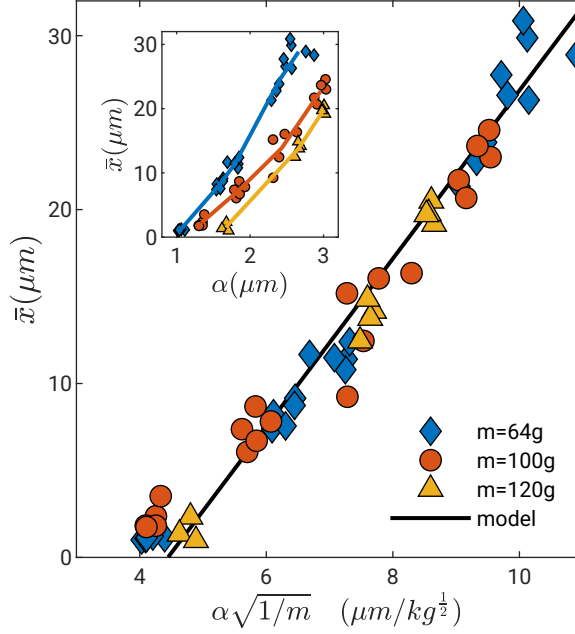
### 2.3. Results and Discussion

To understand the nature of this dynamical system, we conducted initial experiments in which we measured the levitation displacement of a free floating mass placed on top of a vibrating surface in a liquid container, in a fashion similar to other experiments from literature [33–35]. To find the actual capacity of liquid squeeze films to levitate objects stably, we tested the system under ultrasonic sinusoidal excitation of frequency 40 kHz at different vibration amplitudes. As demonstrated in Fig.2.2(a) for a mass of (64 g), experiments confirmed the existence of stable squeeze-film levitation in liquid. To further investigate whether cavitation bubbles play a role in this phenomenon in light of the hypothesis of Tamura et al. [10], we visualized the surface area of the liquid film subjected to the sinusoidal excitation and we found no trace of bubbles in our experiments using a purified liquid.

By looking closely at the response shown in Fig.2.2(b), we observe the following: the reflector reaches an equilibrium levitation position while oscillating in response to the input excitation. We notice that the amplitude of the reflector oscillations is considerably smaller than the input amplitude ( $\approx 15\%$ ), indicating that the reflector oscillations has a minor influence on the squeeze film pressure compared to the input vibration. In addition, we also find the phase shift between the reflector oscillations and input vibrations to be around  $(\frac{\pi}{2})$ , consistent across experiments. Finally, the dynamic response of the system has two characteristic time scales: (a) a time corresponding to the mass reaching a steady-state levitation position  $t_{ss}$  (b) a time associated with the reflector oscillations around its equilibrium position  $t_{fl}$ , such that  $(t_{ss} \gg t_{fl})$ . Since, we are mainly interested in the steady-state time-averaged levitation, we can exploit this observation to decompose the response into two components; namely, a time-averaged and an oscillating components as follows:

$$x = \bar{x} + \tilde{x} = \bar{x} + \gamma \cos(\omega t + \phi) \quad (10)$$

where  $\bar{x}$  is the time-averaged steady-state levitation distance and  $\tilde{x}$  is the oscillation component such that  $\tilde{x} = \gamma \cos(\omega t + \phi)$ . A similar decomposition was previously proposed by Wang et al.[35]. However, in that work, the decomposed parameter was the squeeze film force, not the levitation displacement. By substituting Eq.10



**Figure 2.3. Model validation.** validation of the model demonstrated by the steady-state time-averaged levitation height  $\bar{h}$  as a function of the vibration amplitude and mass of the reflector expressed as  $\alpha\sqrt{1/m}$ . The experimental data were collected using three different masses: 64 g (blue parallelograms), 101 g (orange circles) and 122 g (yellow triangles) at vibration amplitudes up to  $3\mu m$ . The model (black solid line) shows close agreement with the experimental data (individual plotted points) at different combinations of vibration amplitudes and masses. We identified the initial film thickness  $h_0$  to be  $\approx 22\mu m$  and the pressure loss coefficient  $C_e$  to be  $\approx 9$ . The inset shows the decomposed experimental data; the steady-state time-averaged levitation height  $\bar{h}$  as a function of the vibration amplitude  $\alpha$  for each mass individually.

into Eq.8, we can find new expressions for the film thickness  $h(t)$  incorporating the decomposition components. Since we are mainly interested in the time-averaged levitation component  $\langle \bar{x} \rangle$ , we can impose a time average operator  $\langle \cdot \rangle$  on the dynamic equation Eq.9 to find the steady-state time-averaged components. This time average operator is given by  $\langle \cdot \rangle = \frac{1}{T} \int_0^T \cdot dt$ , where  $T$  is the period of the oscillation. To be able to compute the analytical time average of the components of Eq.9, it is essential that the integral of these components using the time average operator to exist. This turns out to be true in the special case when the mass ratio  $M$  is negligible compared to the  $\frac{2}{5\bar{h}}$  term in the denominator of the right hand side of Eq.9. To find the range of values where the assumption of a negligible mass ratio is valid, we carry out an order of magnitude study: for masses of few hundred grams  $O(10^{-1})$ , radii of tenth of millimeters  $O(10^{-2})$  and for a typical liquid density of order  $O(10^3)$ , we find the



order of magnitude of the mass ratio to be  $O(10^3)$  compared to an  $O(10^5)$  for the  $\frac{2}{5h}$  term. Therefore, we can safely conclude that the assumption we made priori (i.e.  $M \ll \frac{2}{5h}$ ) is valid for the aforementioned ranges of values of mass, radii and density. By imposing this assumption on Eq.9, we obtain the following expression:

$$\ddot{x} = \frac{25\dot{h}^2}{14h} - \ddot{y} - \frac{10\nu\dot{h}}{h^2} - \frac{5C_e\dot{h}^2}{8h} - \frac{5hMg}{2} \quad (11)$$

A remarkable consequence of the  $M \ll \frac{2}{5h}$  assumption is that all of the terms of Eq.9 become symmetric in Eq.11 except for the convection terms  $\frac{25\dot{h}^2}{14h}$  and  $\frac{5C_e\dot{h}^2}{8h}$ . It means that within the range of values in which the assumption is valid, the convection effects prevail and becomes the sole source of the steady-state time-averaged levitation. By imposing the time average operator, the symmetric terms  $\ddot{x}, \ddot{y}$  and  $\frac{10\nu\dot{h}}{h^2}$  of Eq.11, by definition, converge to zero which yields the following:

$$\langle \dot{h}^2 \rangle = \frac{(20 - 2.2C_e)}{28Mg} \langle \dot{h}^2 \rangle \quad (12)$$

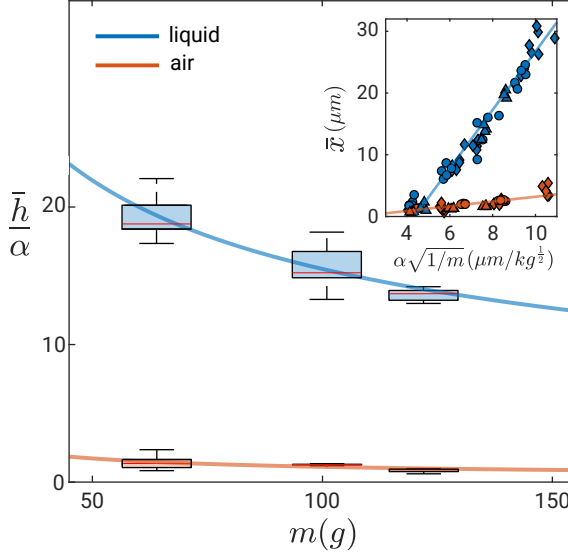
The time average of the film height yields the steady-state component  $\bar{h}$  which is the main variable that we are trying to find here. On the other side of the equation, we can find the time average of the  $\dot{h}^2$  by substituting the derivative of Eq.10 into the  $\dot{h}$  expression of Eq.8 and find the time average integral, to be:

$$\langle \dot{h}^2 \rangle = \frac{\alpha^2\omega^2}{2} + \frac{\gamma^2\omega^2}{2} - \alpha\gamma\omega^2 \cos(\phi) \quad (13)$$

Given our earlier experimental findings that the amplitude of the reflector oscillations is typically ( $\gamma \leq \frac{\alpha}{6}$ ) and that the phase shift is consistently ( $\phi \approx \frac{\pi}{2}$ ), we can conclude that the reflector dynamics terms  $\gamma^2\omega^2/2$  and  $\alpha\gamma\omega^2 \cos(\phi)$  are negligible compared to the vibration input term  $\alpha^2\omega^2/2$ . By omitting the negligible terms from Eq.13, substituting it back in Eq.12 and expanding the mass ratio  $M$ , we find the following expression for the time-averaged steady-state film height  $\bar{h}$ :

$$\bar{h} = \pi R^2 \alpha \omega \Phi \sqrt{\frac{\rho}{mg}} \quad (14)$$

where  $\Phi(C_e)$  is a correction factor that encapsulates the energy dissipation due to the edge effect, such that  $\Phi = \sqrt{(20 - 2.2C_e)/56}$ . The derived physical law reveals the fundamental relationship between the steady-state time-averaged levitation height



**Figure 2.4. Comparison of the load-carrying capacity between the in-liquid and in-air squeeze-film levitation.** The load-carrying capacity is demonstrated by the box-plot of the amplification factor  $\frac{\bar{h}}{\alpha}$  as a function of the mass of the floating reflector where the solid lines are power law regression. The comparison shows that at the same mass, the in-liquid system achieves a significantly greater amplification than the levitation in air. The inset shows the steady state levitation distances  $\bar{x}$  of the in-liquid and in-air levitation at different combinations of vibration amplitudes and masses, indicating that both incompressible liquids and air follow a linear relationship with  $\alpha\sqrt{1/m}$ , but with different slopes, as expected from our model for the liquid case and the literature for the air case.

and the system parameters in the case of prevailing convection effects. In comparison to the derived formula by Wang et al.[35] for in-air levitation, we see that  $\bar{h}$  scales linearly in both cases with the vibration amplitude  $\alpha$  and with the inverse of the square root of the mass  $\sqrt{1/m}$ . By contrast, in the liquid case,  $\bar{h}$  is proportional to the surface area  $\pi R^2$ , the angular frequency  $\omega$  and the square root of the liquid density  $\sqrt{\rho}$  while in the air case,  $\bar{h}$  is proportional to the square root of the surface area  $\sqrt{\pi R^2}$  and has no direct relation with the angular frequency nor with the density. These differences are well justified by the fundamental difference between the working mechanism of in-air (compressibility) and that of in-liquid systems (inertial).

To validate the theory, we conducted experiments using different combinations of masses (64, 101 and 122 g) with a disk of radius (10 mm) and ultrasonic (40 kHz) vibration amplitudes (1-3  $\mu m$ ), in a purified liquid of density (1030 kg/m<sup>3</sup>). Given the relative scaling between the surface area  $\pi R^2$  and the input vibration angular frequency  $\omega$  in Eq.14, we notice that for our disk size  $O(10^{-2})$ , it was necessary to have an ul-

trasonic excitation of angular frequency  $\omega \sim O(10^5)$  to obtain a measurable levitation height.

Using the collected experimental data, we identified the empirical parameters  $(h_0, C_e)$  to be  $h_0 \approx 22\mu m$  and  $C_e \approx 9$ . As demonstrated in Fig.2.3, the model shows close agreement with the experimental data. We also observe that the correction factor  $\Phi$  is not influenced by the vibration amplitude nor the mass, as expected, since the loss coefficient, by definition, depends only on the geometry of the film entrance. In addition, it is interesting to see that levitation starts to occur at certain amplitude threshold which mainly stems from the initial gap thickness  $h_0$ , as expected from our model. The observation was also noted for air squeeze films [35]. This effect is mainly because for thicker squeeze films, higher amplitude is required to induce sufficient pressure in the film to outweigh the mass of the reflector and vice versa.

Since air squeeze levitation systems is the de facto standard in the literature, we conducted the same set of experiments in air. As demonstrated in Fig.2.4, we see that the amplification factor  $\frac{\bar{h}}{\alpha}$ , which is the ratio of the steady-state levitation height to the amplitude of the input vibration, is significantly higher in liquid compared to air across different masses, reaching levitation amplification ten times larger in the liquid case. We also observe that while both liquid and air scales linearly with  $\alpha\sqrt{1/m}$ , the slope in both cases is different. The slope in the liquid case is around ten times greater than the air case. The difference in slope demonstrates the significantly greater load-carrying capacity of liquid squeeze films compared to air, based solely on the squeeze film effect.

## 2.4. Conclusion and Future Work

In summary, we showed that the liquid inertia, manifested as the temporal and convective acceleration of the liquid film, is the source of pressure asymmetry that causes the levitation phenomenon to exist. We demonstrated that in the special case of a negligible mass ratio ( $M$ ) compared to a film thickness parameter ( $\frac{2}{5h}$ ), the fluid convective inertial effects prevail and become the sole source of the levitation phenomenon. In such a case, we simplified the dynamics equation and we derived a closed form compact formula that predicts the steady-state time-averaged levitation height as a function of the vibration input parameters  $(\alpha, \omega)$ , liquid density  $(\rho)$  and mass  $(m)$  of the levitated object. Lastly, we conducted a comparative experiment

of load carrying capacity of liquid and air squeeze film levitation systems, demonstrating a significantly greater load-carrying capacity for in-liquid systems. The findings of this letter could potentially lead to novel physics-informed designs of in-liquid levitation systems for non-contact manipulation, transportation and friction modulation applications.

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# CHAPTER 3

## Ultrasonic Lubrication for Variable-Friction Catheters

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## Abstract

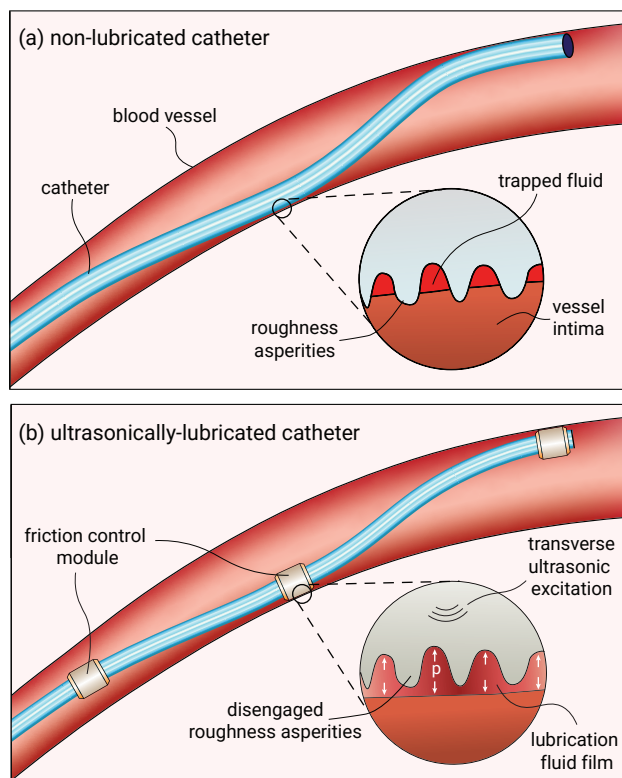
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During minimally invasive endovascular procedures, catheters are guided through blood vessels to perform interventions such as cutting or suturing. In these procedures, friction between the catheter and the vessel wall is unavoidable and plays a dual role: while it can enhance stability during the intervention, it also risks damaging the inner layer of the vessel wall during navigation, potentially leading to post-operative complications such as infection and thrombus formation. Building on the physics foundation established in Chapter 2, this chapter investigates the use of ultrasonic lubrication to develop variable-friction catheters capable of switching on demand between low- and high-friction states, to facilitate navigation and restore positional stability when needed, respectively. In this chapter, we develop and evaluate a proof-of-concept friction-control module based on ultrasonic lubrication as the fundamental building block of the variable-friction catheter design. Our experiments demonstrate that the prototype can effectively reduce friction by up to 11% and 60% on average, on soft and rigid surfaces, representing its potential performance on healthy and calcified tissue, respectively. These results demonstrate the viability of ultrasonic lubrication as an active friction-control strategy for endovascular catheters, translating the physical principles established in Chapter 2 into a functional component for catheters and providing the basis for the broader integration and ex vivo tissue validation discussed in Chapter 4.

### 3.1. Introduction

Over the past few decades, minimally-invasive endovascular interventions superseded conventional open heart surgeries, leading to shorter recovery times and lower infection rates. In a typical endovascular catheterization procedure, the interventionist inserts a catheter in the radial or femoral artery and navigates it through the arteries to the heart, where the intervention is performed. In order to safely reach the heart, catheters (and guidewires) used during these procedures need to easily follow the curves in the vascular system, while creating as little friction as possible with the blood vessel wall. When these devices exhibit high friction, their insertion risks damaging the mucous membranes or the intima of the blood vessels, which may lead to infectious diseases or thrombus formation [2–4]. While low friction is beneficial to avoid damage to the membranes and blood vessel wall, it complicates holding a specific location in open spaces, such as inside the heart. This problem is particularly true when high forces need to be applied, such as when cutting or puncturing tissues. This suggests the need for an active friction management technique that can regulate the frictional properties of catheters depending on the phase of the catheterization procedure; for instance, having low friction while navigating through the vasculature and switching to a high friction state while executing the surgical task.

Friction management strategies in catheterization procedures are typically passive. Current practices involve direct application of lubricants, aiming to reduce the risk of tissue damage during sliding [5–8]. Another approach uses specially coated self-lubricating instruments, like hydrophilic catheters, which become lubricated when into contact with a wet surface [9, 10]. The first approach, involving direct application of lubricants, faces a limitation: the effectiveness of the lubricant is short-lived and easily wiped off during sliding. Consequently, its utility is restricted to relatively short distances [6]. The second approach, centered around specialized coatings, like hydrophilic surfaces, is procedure-specific, primarily applicable to procedures involving water-based fluid contact such as urology, and lacks versatility for broader catheterization applications. Furthermore, the inability to switch between high and low friction levels in both methods poses challenges in instrument stabilization. The limitations of the existing practices evidently suggest the need for an active friction modulation strategy. A strategy that allows instruments to transition between high and low friction states in real-time based on the different stages of the intervention.



**Figure 3.1. Ultrasonically-lubricated catheters.** (a) conventional catheter in a blood vessel where contact occurs at multiple points along its lumen. The magnified view shows the asperities contact at a microscopic scale, which results in complications when the catheter slides relative to the vessel. (b) the proposed variable-friction catheter with integrated friction control modules along its shaft. The magnified view shows the interface of the friction control module surface and the vessel, which creates an active lubrication fluid film resulting in a disengagement of the asperities leading to a smooth and safe sliding.

We propose a catheter design in which friction can be programmed. The catheter is built around several discrete modules each of which can provide friction modulation using ultrasonic vibration along the shaft of the catheter. In this chapter, we present a proof-of-concept of the friction control modules, alongside results from simulation and experimental validation, including sliding friction experiments.

## 3.2. Working Principle and Design

### 3.2.1. Ultrasonic Lubrication

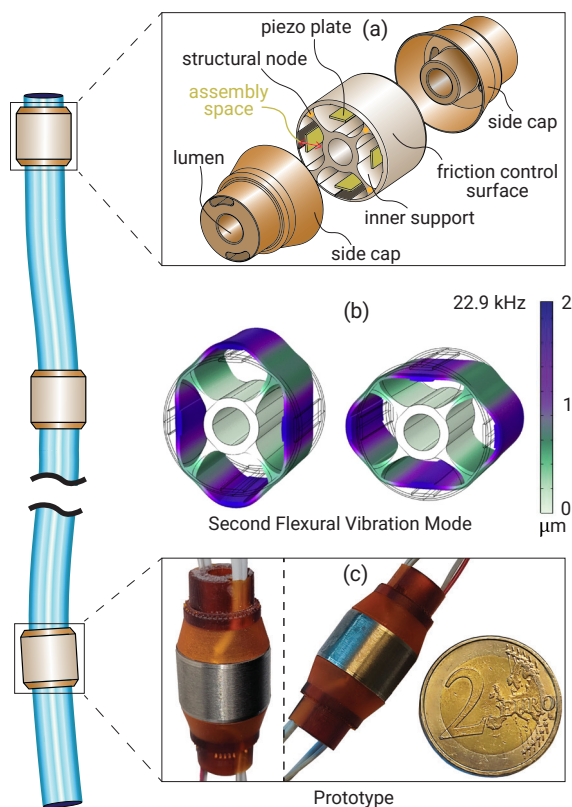
When two objects come into contact and undergo relative sliding, friction arises as a consequence of the interaction between a finite number of microscopic surface roughness asperities, as illustrated in Fig.3.1(a). These asperities form junctions with the opposite surface that collectively form a *real* area of contact, which is significantly smaller than the apparent area of contact. Consequently, a substantial portion of the apparent contact area remains disengaged, entrapping fluid from the surrounding medium and forming a thin film of fluid, known as a squeeze film, along the interface.

When transverse ultrasonic vibrations are applied to one of the objects in contact, the fluid trapped between the asperities undergoes non-linear compression, resulting in a net positive overpressure in the fluid film. This overpressure causes the two objects to separate, as shown in Fig.3.1(b), resulting in a controllable coefficient of friction—a phenomenon known as ultrasonic friction modulation or ultrasonic lubrication [11, 12]. The amount of separation is monotonically correlated to the vibration amplitude, enabling precise and controllable active lubrication at the interface [11].

Ultrasonic lubrication technology has demonstrated its effectiveness across various applications, particularly in scenarios where friction modulation is essential, such as in tactile interfaces [13], contactless manipulation [14] and squeeze-film bearings [15, 16]. Moreover, ultrasonic lubrication has proven its capability to function in submerged liquid environments [17] as effective as in air environments, showcasing its versatility to different operating conditions. The capacity of ultrasonic lubrication to seamlessly transition between high and low friction states in different environments suggests its potential to fulfill the evident need of active friction modulation in catheters.

### 3.2.2. Conceptual Design of Variable-Friction Catheters

Leveraging the concept of ultrasonic lubrication to actively control friction in catheters, we introduce a discrete module design concept in which friction control modules are integrated along the shaft of the catheter as shown in Fig.3.2. The design requirements for the friction control modules, to ensure proper functionality, includes technology and application requirements. To be effective, the modules should achieve ( $\geq 2 \mu\text{m}$ ) of vibration amplitude at an ultrasonic frequency ( $\geq 20 \text{ kHz}$ ) for maximal



**Figure 3.2. Design, simulation and implementation of friction control modules.** (a) shows the design of the friction control modules featuring a ring structure and two side caps. The ring structure encompasses four piezoelectric plates and an inner support structure attached to the outer surface at the nodal lines to minimize vibration interference. (b) shows the finite element simulation of the vibration mode of the designed ring structure. (c) shows the implemented scaled-up prototype of the designed resonating ring structure.

lubrication [18, 19]. The catheter application requires the design to be tubular, to have a lumen and to have the potential to be downsized. With these requirements in mind, we designed friction control module, illustrated in Fig.3.2(a), which features a resonating ring structure and sealing side caps. The resonating ring encompasses a vibrating surface and an inner support structure with a lumen. The vibrating outer surface of the resonating ring comprises four piezoelectric plates, as shown in Fig.3.2(a), such that when the piezoelectric plates are activated, a standing flexural wave is induced in the ring as depicted in Fig.3.2(b). The frequency of this wave coincides with the second flexural resonance mode (second harmonic) of the ring structure, which is governed by the diameter and wall thickness of the ring. To properly support the

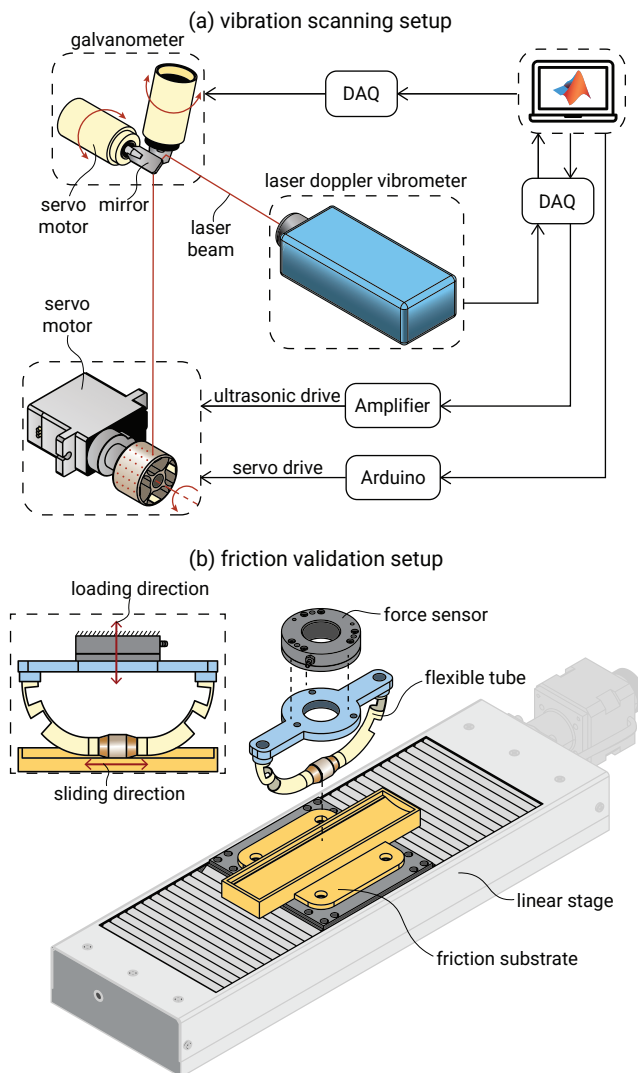
ring, the inner support structure is connected to the ring at the nodal lines of the standing wave through thin walled webs, as illustrated in Fig.3.2(a). These webs ensure minimal interference with the generated wave. The ring structure is sealed from both sides through the side caps, which insulates the piezoceramic plates chambers while featuring routing for the power wires. The wall of the side caps are designed to be flexible to not interfere with the ring vibrations.

### 3.2.3. Design and Simulation of Friction Control Modules

We carried out a computational finite element method (FEM) simulation to estimate the resonant frequency and the potential amplitude that this arrangement can reach. We first selected a scaled-up size for the concept with a diameter of 15 mm for ease of manufacturing. We built the model using COMSOL Mutliphysics® (COMSOL AB, Stockholm, Sweden) in which we carried out two types of studies. We first ran an eigenfrequency study to identify the resonant frequency corresponding to the desired vibration mode. Then we ran a frequency response study in which we simulate the behavior of the ring structure when the piezoceramic plates are activated at a given frequency, which we use to estimate the vibration amplitude and mode shape. The simulation process iterated on the design parameters to find the combination that would result in an ultrasonic frequency and vibration amplitude that meets the design requirements. For the design parameters selected (titanium, diameter of 15 mm, wall thickness of 0.5 mm), we found the second flexural resonant frequency to be (22.9 kHz) and the amplitude to potentially reaching (2  $\mu$ m), as shown in Fig.3.2(b).

### 3.2.4. Prototype Development

We fabricated a scaled-up titanium ring prototype with a diameter of 15 mm, wall thickness of 0.5 mm and length of 10 mm using wire discharge machining. A set of four 1 W ceramic piezoelectric plates, 0.3 mm thick, were then cut into the right size and glued manually to the fabricated ring using a specially-designed wedge system to apply a uniform force along its length. Afterwards, the 0.5 mm power wires were soldered and routed through the assembled side caps. The exploded view of the prototype assembly is illustrated in Fig.3.2(a).



**Figure 3.3. Experimental setups.** (a) experimental setup to characterize the friction control module performance in terms of resonant frequency, maximum vibration amplitude, vibration mode and effective surface area. (b) experimental setup to validate the capacity of the friction control modules in modulating friction on soft and rigid substrates, representing healthy and calcified tissue, respectively.

### 3.3. Experimental Validation

To validate the concept of the friction control modules and characterize its performance, we conducted two experiments. The first experiment aimed to characterize the vibration performance of the resonating ring. The second experiment aimed to

validate the friction control modules capability to modulate friction on different type of surfaces.

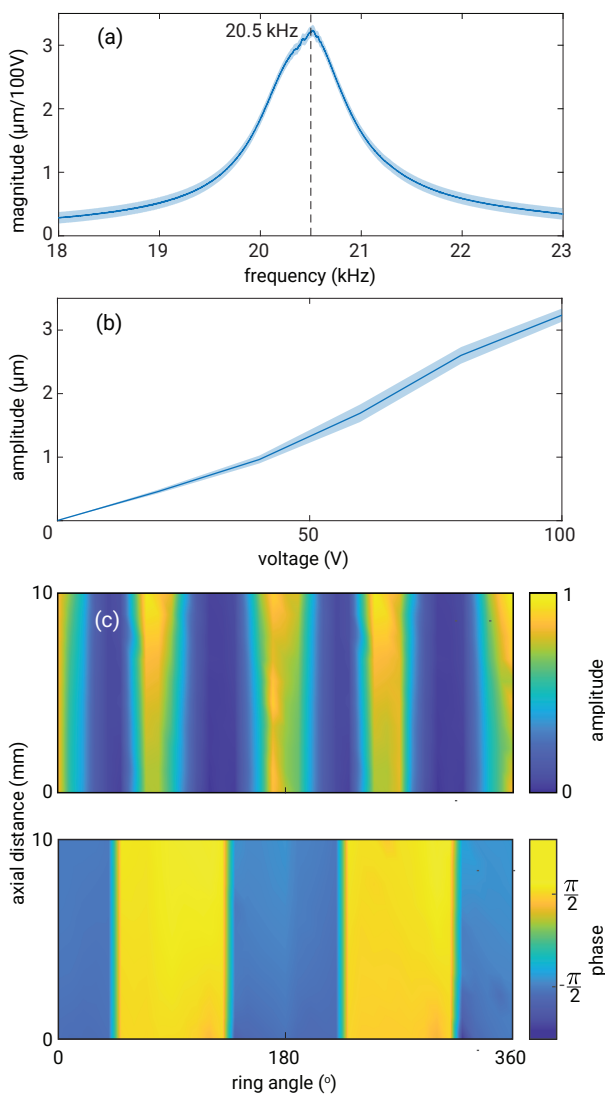
### 3.3.1. Vibration Characterization Experiment

In this characterization experiment, we measured the vibration response of the resonating ring to the voltage input using a single point laser doppler vibrometer (LDV) (OFV5000, Polytec GmbH, Baden-Württemberg, Germany). To be able to scan the whole surface of the ring, we affixed the ring to a servo motor to allow the ring to rotate while controlling the laser beam of the LDV using a dual axis galvanometer (ScannerMAX Saturn 5B, Edmund Optics Inc., NJ, U.S.A.), as shown in Fig.3.3(a). The galvanometer scans the ring longitudinally while the servomotor scans the ring at different orientation angles. The control of the galvanometer, servo motor and the readings of the LDV was facilitated through data acquisition cards, controlled through a MATLAB interface.

To quantify the performance of the ring, we employed four key evaluation criteria: the resonant frequency, the maximum vibration amplitude, the vibration mode and the effective surface area. We define the effective area as the area in which the vibration amplitude reaches more than or equal to two microns, essential to fulfill the technology requirements. For this quantification, we do three different measurements; a frequency response measurement to quantify the resonant frequency, a voltage sweep measurement at the identified resonant frequency to quantify the maximum amplitude and a surface scanning of the ring to identify the resonance mode and quantify the effective surface area.

The experimental procedure initiated by collecting the frequency sweep at the central point of each of the four anti-nodal lines followed by the voltage sweep at the identified resonant frequency. Afterwards, the surface scanning procedure started by positioning the ring at an anti-nodal line intersecting the laser beam path. Subsequently, we identified and discretized the measurable path into ten points longitudinally. The measurement algorithm then activated the transducers, concurrently recording responses at each designated point. Upon completion of one line scan, the ring underwent a 5-degree rotation, and the process iterated until a comprehensive response dataset around the ring was obtained.





**Figure 3.4. Vibration characterization of the friction control module.** (a) frequency response of the resonating ring along the anti-nodal lines. The solid line is the mean of the measurements across the four anti-nodal lines and the shade represents the standard deviation. (b) mapping between the input voltage and the output vibration amplitude along the anti-nodal lines. The solid line represents the mean of the measurements across the four anti-nodal lines and the shade represents the standard deviation. (c) vibration surface map of the resonating ring. The top color map represents the normalized vibration amplitude and the bottom shows the corresponding phase map. The measured surface map closely matches the anticipated vibration mode from simulation shown in Fig.3.2(b), validating the design.

### 3.3.2. Friction Validation Experiment

To validate the capacity of the friction control modules for modulating friction, we conducted sliding friction experiments on both rigid and soft surfaces. In this study, we secured the ring structure to a 6-axis force sensor (Nano43, ATI Industrial Automation, NC, U.S.A.) using a pair of flexible tubes to emulate the stiffness force of catheters. The module-sensor assembly was affixed to a manual linear slider to regulate the normal load. The friction substrates, affixed opposite to the surface of the resonating ring, were allowed to slide against the ring surface using a motorized linear stage. Simultaneously, the LDV tracked the vibration amplitude from top of the resonating ring while sliding.

Friction experiments were conducted on two different friction substrates. The first one was a rigid 3D printed flat substrate using a rigid resin material (Model, FormLabs, MA, U.S.A.) with a post-cured elastic modulus of 2.3 GPa. The second substrate was a soft 3D printed flat substrate using a soft resin material (Elastic, FormLabs, MA, U.S.A.) with a post-cured ultimate tensile strength of 3.23 MPa and shore hardness of 50A. These two substrates represent the two ends of the mechanical properties spectrum in arteries: soft material representing healthy arteries and rigid material representing calcified arteries. Testing on both materials is essential because, while arteries are in principle soft, in reality, > 70% of healthy individuals who are 40 – 49 years old suffer, to varying degrees, from coronary atherosclerosis [20–22], which involves the buildup of hard calcified plaque in arteries. Patients undergoing catheterization procedures already suffer from an advanced form of atherosclerosis, making it more likely for catheters to encounter calcified regions. This calcification stiffens the artery in general and more importantly on a local level, it forms rigid interface with catheters which was shown to have significantly higher coefficient of friction compared to healthy tissue [23]. Such a high friction with the calcified plaque can dis-integrate particles of the calcified plaque, leading to adverse consequences as they migrate through the bloodstream. Additionally, testing on rigid surfaces allows for comparing the performance of the proposed design to the existing devices in the benchmark application of haptic touchscreens [13, 18].

The experimental protocol began with loading the resonating ring on the friction substrate to (1 N), a critical load for catheters [24]. Then, the substrate started to slide at 1 mm/s for a distance of 15 mm. Two set of experimental data were collected. The first one was for a proof-of-concept test in which we activated the ultrasonic vibrations five seconds after sliding started, for a period of five seconds at a constant

maximum vibration amplitude. In the second test, we activated a linearly modulated input voltage amplitude from a minimum to a maximum for a period of five seconds while sliding. Each sliding friction test was repeated six times.

To quantify the friction modulation capability, we use the friction reduction percentage, representing the amount of reduction in coefficient of friction as compared to the initial coefficient of friction before activating the vibrations, which can be calculated as follows:

$$\text{friction reduction (\%)} = \frac{|\mu_{\text{off}} - \mu_{\text{on}}|}{\mu_{\text{off}}} \quad (3.1)$$

### 3.4. Results and Discussion

The experiments conducted to characterize the vibration of the resonating ring prototype unveiled a resonant frequency of approximately 20.5 kHz as shown in Fig.3.4(a), slightly shifted from the frequency predicted from simulation. Additionally, we found the maximum vibration amplitude at the anti-nodal lines to be  $\approx 3.2 \mu\text{m}$ , as shown in Fig.3.4(b), surpassing the predetermined amplitude of the design requirements. The observed variance between the experimentally identified resonant frequency and the simulation-predicted value can be attributed to discrepancies in the material properties between the predefined titanium alloy used in simulation and the actual alloy used to fabricate the prototypes. Additionally, the model assumes perfectly homogeneous material properties and does not account for the effects of machining processes and tolerances on the prototypes.

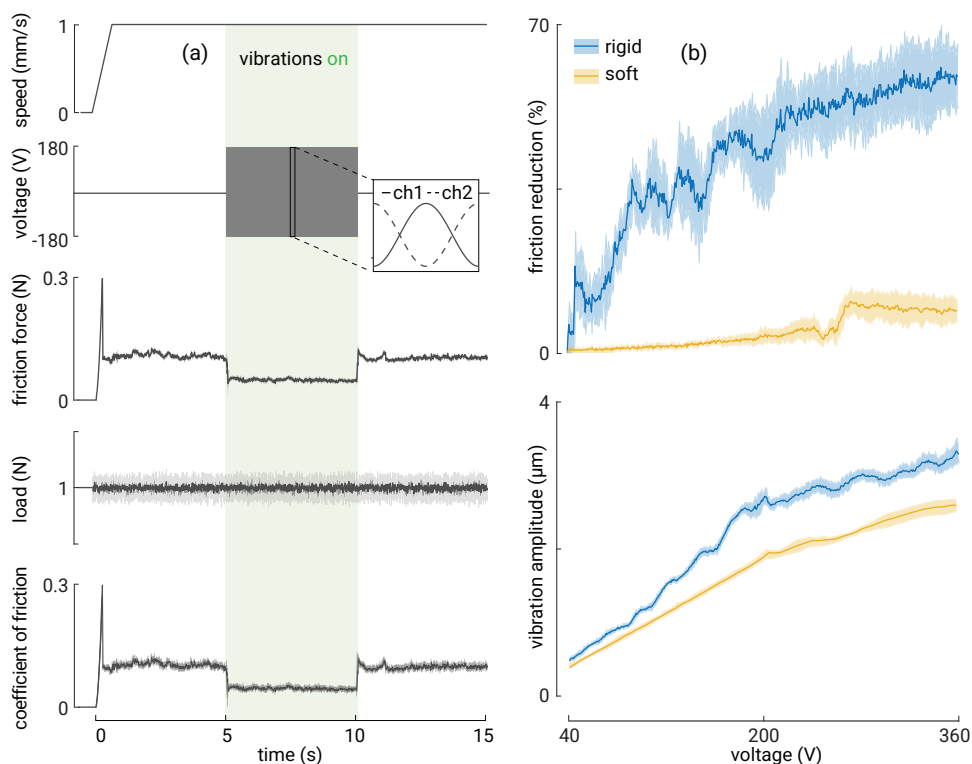
Upon scanning the surface of the resonating ring, at the identified resonant frequency, the corresponding mode was validated to be the second flexural mode as shown in Fig.3.4(c), which closely matches the anticipated mode from simulation. The surface map further revealed that the active vibration surface of the resonating ring is approximately  $\approx 50\%$  of the entire surface of the resonating ring, implying that the friction modulation capability is limited to only half of the ring surface area. This relatively small active surface can be attributed to the rigid support connection at the nodal lines which limits the nodal line ability to rotate, thus limiting the vibration amplitude in the neighbouring areas.

The proof-of-concept friction experiment demonstrated the impact of activating the friction control module on the friction dynamics. Upon activating vibration, at the 5

second mark in Fig.3.5(a), an immediate reduction in both friction force and the coefficient of friction by  $\approx 60\%$  was observed. Subsequently, upon deactivation at the 10 second mark (Fig.3.5(a)), the friction values reverted to their original state, confirming the friction control module ability to reduce friction. This demonstrated range of friction reduction matches the 60-80% typical friction reduction achieved by ultrasonic haptic touchscreens [13, 18], which serve as the benchmark for ultrasonic lubrication technology. This experiment was conducted on a flat surface interfacing with the curved surface of the friction control module, resulting in a line contact with very limited overlapping surface area. The overlapping surface area plays an important role in generating the desired lubrication effect, and despite it being limited, the module still exhibited significant friction reduction upon activation, showcasing its potential. This conclusion can be explained by the well established fact that friction force does not depend on the contact area. Therefore, the increase in contact area, associated with the increase in curvature, resulted in increasing the squeeze-film force but not the friction force, resulting in a net increase in the friction reduction capacity.

The voltage modulation friction experiment showed a corresponding modulation of the vibration amplitude, which in turn resulted in friction modulation at the interface as shown in Fig.3.5(b). In the rigid case, the amount of friction reduction changes gradually across the entire input range reaching up to  $\approx 60\%$  reduction on average, showcasing the system ability not only to toggle between high and low friction states but also to finely control friction within this range. We observe a measurable friction reduction that only starts to appear at a certain vibration amplitude  $\approx 0.6 \mu m$ . This value can be explained by the squeeze-film theory since a given amount of vibration amplitude is required to build sufficient pressure and be able to separate the two surfaces. We also observe that once the vibration amplitude exceeds the working threshold, the friction reduction percentage increases rapidly to the 35% level and afterwards it follows a linear trend towards the maximum reduction level. A possible explanation for the high sensitivity at the initial region can be that once the pressurized lubrication layer is formed, the disengagement of the asperities deep into the contact results in a high reduction gain. Once the shallow asperities are levitating, the system behaves more linearly.

Similarly, on the soft surface, modulating the voltage had a direct effect on the vibration amplitude. However, the friction reduction percentage appears to have significantly different dynamics as shown in Fig.3.5(b). At first, we observe that no notable reduction in friction is achieved while the amplitude continues to increase up to  $\approx 2 \mu m$ . Beyond this point, the reduction starts to increase linearly before rapidly



**Figure 3.5. Results of the friction validation experiments.** (a) results of the proof-of-concept experiment which show that upon turning the vibrations on, both the friction force and the coefficient of friction decrease immediately by  $\approx 60\%$ , and revert promptly to its original state once vibrations are turned off. The observed behavior validate the impact of the proposed design for reducing friction. The solid lines represent the mean and the shadows represent the standard deviation of the repetitions. (b) results of the voltage modulation experiment on soft and rigid substrates, representing healthy and calcified tissue, respectively. In the rigid substrate case, the modulation of voltage modulated the vibration amplitude, resulting in linearly controllable friction, reaching up to  $\approx 60\%$  of friction reduction. In the soft substrate case, the voltage modulation similarly induce a modulation of the vibration amplitude. However, the friction reduction is only detectable at higher vibration amplitude, after which the reduction percentage follows a linear trend before saturating at  $\approx 11\%$ .

saturating at  $\approx 11\%$  on average. This behavior is well explained by the elasticity of the soft surface which, at the beginning, absorbs energy from the vibrating surface, preventing the lubrication layer from developing. Once vibration reaches a certain threshold, a separation between the two surfaces occurs and the lubrication layer can develop. The inherent elasticity, however, continues to play a role in limiting the pressure of the lubrication layer which thus causes the reduction to saturate.

It is important to note that the results obtained on the soft substrate may not accurately reflect what we would observe when testing on soft tissues, as these results are

more specific to the damping properties of the soft material used in our tests. Previous research by Friesen et al. [25, 26] demonstrated that the damping factor of soft materials significantly affects the performance of ultrasonic lubrication. This significance of damping is because ultrasonic lubrication systems produce extremely low amplitudes, on the order of  $O(10^1) \mu m$ , which results in minimal elastic effects, while vibration velocities are relatively large, on the order of  $O(10^2)$  cm/s, leading to significant viscous damping effects. Moreover, Friesen et al. [25] found that materials with high damping factor is more favorable because its damping delays the response of the soft material to the source vibration, resulting in a phase shift between the vibration source and the soft surface. This phase shift allows a pressurized squeeze film to develop at the interface, leading to physical separation between the two surfaces and thus reduction in friction. Conversely, materials with low damping follow the vibration profile of the vibration source, preventing the squeeze film from developing and thus resulting in no friction reduction. In their comparative experiment [26], they also found that human skin exhibited significantly higher damping compared to other common soft materials such as DragonSkin silicone rubber, similar to the material we used, indicating a potentially better response of actual soft tissues to ultrasonic lubrication.

One implicit assumption in our experiments is that contact occurs exclusively between the friction control modules and the friction substrate. While this is useful for assessing the performance of the friction control modules, in a real-world scenario, the main body of the catheter may also contact the blood vessel wall. Ideally, the modules should be slightly larger than the catheter's diameter to ensure that contact with the blood vessels occurs primarily with the friction control modules, minimizing contact with the catheter body itself. Thus, the size of the modules relative to the catheter diameter, as well as the distance between the modules, is crucial. Additionally, the catheter body should be coated with an ultra-low friction material to ensure smooth interaction with the blood vessel if contact occurs between the catheter body and the tissue. This ultra-low friction coating becomes feasible in this case since the friction control modules will provide stability when necessary by switching to their high-friction state.

One limitation of our current prototype is that the wires powering the piezoelectric actuators are routed through the lumen, obstructing it, and potentially limiting the catheter's functionality by reducing the available space for tool insertion. This issue can be addressed by embedding the wires within the catheter body, as demonstrated by Abdelaziz et al. [27], who developed a methodology to embed wires of various

sizes and customized arrangements into the catheter body, enabling seamless power transmission throughout the catheter without compromising its functionality.

Other limitations of our current prototype are the difficulty to downsize and its limited lumen utilization,  $\approx 20\%$ , representing the lumen diameter compared to the overall module diameter. Both limitations arise from the need to maintain sufficient space between the piezoelectric plate and the inner support structure, illustrated as assembly space in Fig.3.2(a), to facilitate the assembly of the plates. One potential solution is to have a separable inner support from the outer structure of the friction control module. This design would allow the inside of the module to be completely free during the assembly of the piezoelectric plates, providing more room for their installation. After assembly, the inner support could be inserted into place without requiring additional space for the assembly process, thereby maximizing lumen utilization post-assembly and enabling the possibility of downsizing. By designing this separable inner support to touch the nodal lines with needle-like edges, we could potentially also overcome the limitation of the small active vibratory area discussed earlier. This design would allow the nodal lines to rotate freely around the needle edges, unlike the rigid connection in the current design, thereby enhancing performance.

### 3.5. Conclusion and Future Work

We introduced a novel concept of a variable-friction catheter comprising discrete active resonating modules along its shaft. These modules use ultrasonic lubrication to actively control the coefficient of friction of its surface when in contact with the environment. We designed and implemented a scaled-up prototype of the friction control modules to validate our design. The experimental characterization showed that the implemented prototype fulfilled the technology requirements by successfully achieving  $\approx 3.2 \mu\text{m}$  of vibration amplitude at an ultrasonic resonant frequency of  $20.5 \text{ kHz}$ . Vibrometry validated that the resonant mode matches that of the simulation and showed that the active vibratory surface area is  $\approx 50\%$  of the total surface area. Through sliding friction experiment, we validated that the modules can modulate friction on both soft and rigid substrates, representing soft and calcified tissue, with a maximum friction reduction percentage of  $\approx 11\%$  and  $\approx 60\%$ , respectively.

In the future, we plan to improve the miniaturization and to increase the active vibratory surface area. Both improvements are expected to lead to increased friction reduc-

tion particularly in scenarios where the modules are in contact with curved compliant interfaces. Moreover, we will investigate the effect of different parameters including the curvature of the sliding surface and the orientation of the friction control modules. Both parameters are key to understand the friction modulation phenomenon when the interfaces are curved. In addition, we plan to develop a waterproof design to allow for testing in liquid environments, a crucial aspect of the environment in which catheters operate. Furthermore, we plan to test the efficacy of the system in a clinically relevant environment on lubricated ex-vivo healthy and calcified tissue. Finally, we will assess the safety of the technology by capturing the thermal footprint of the friction control modules while in contact with tissue and examining the surface of tissue samples for potential damage.

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# CHAPTER 4

## Active Lubrication of Transluminal Medical Instruments

Under review as: M. A. Atalla, J. Nieuwenhuis, A. Martin, X. Wang, A. Canden, M. J. Carré, R. Lewis, A. Sakes and M. Wiertlewski, '*Active Lubrication of Transluminal Medical Instruments*' [1].

## Abstract

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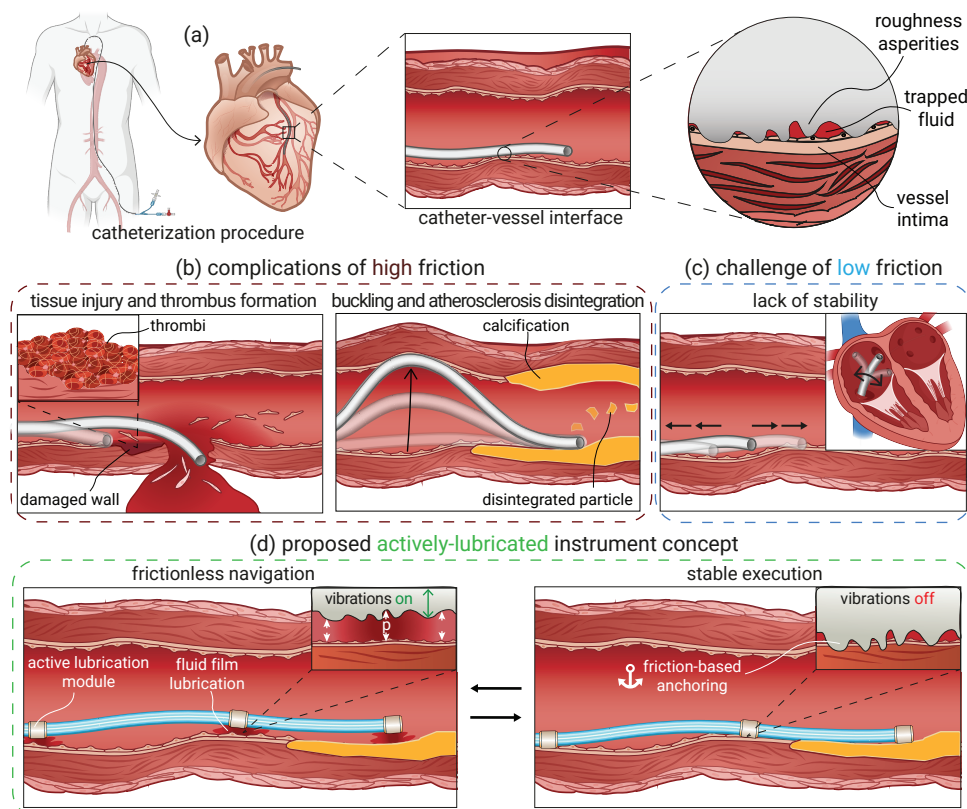
Transluminal minimally invasive procedures use natural orifices or small incisions to access internal anatomical structures, reducing surgical trauma and supporting faster recovery. However, advancing instruments such as catheters and endoscopes through confined anatomical pathways inevitably creates frictional interactions with luminal tissues, which can compromise navigation and increase the risk of complications such as perforation, poor haptic feedback, and instrument buckling. This chapter advances the proof-of-concept friction control modules developed in Chapter 3 to optimize assembly and system integration, successfully miniaturizing the modules down to 10 mm and 3 mm implementations while simultaneously improving vibration performance. Furthermore, this chapter extends the experimental evaluation to examine the influence of interface curvature, lubrication conditions, and interface elasticity on the performance of the active lubrication modules. Crucially, this chapter provides the first *ex vivo* validation of the concept, demonstrating friction reductions of up to 42% on porcine aorta tissue, and examines the thermal safety of the modules for future translational development. This chapter concludes Part I by translating ultrasonic lubrication from foundational physics (Chapter 2) and proof of concept (Chapter 3) into a scalable, functional design of active lubrication system for transluminal medical instruments to 'glide' or 'grip' on demand.

## 4.1. Introduction

Minimally invasive surgeries have reshaped healthcare by reducing patient trauma, shortening recovery time, and improving clinical outcomes. Among these techniques, transluminal interventions, which access internal anatomy through natural orifices and small incisions, enable procedures at difficult-to-reach anatomical targets [2–4]. These interventions typically follow a two-phase structure: first, instruments navigate through anatomical pathways to the target, and second, once the instrument is positioned, surgical tasks are performed at the target site. Because tissue interaction differs markedly between these phases, the design of transluminal instruments must accommodate contrasting functional requirements. In particular, friction must be low during insertion for smooth navigation and minimal tissue damage, yet needs to remain high when the instrument is at the target location for stability.

During navigation, the insertion of transluminal instruments, such as catheters and endoscopes, causes frictional interactions with surrounding tissue [5, 6], as illustrated in Fig. 4.1(a). This friction causes tissue damage, bleeding, and patient discomfort across catheterization [7–12], colonoscopy [13–15], and upper gastrointestinal endoscopy [16–18]. It also diminishes haptic transparency and alters the interventionist's force perception, increasing the risk of accidental dissection or perforation [19–21]. Excessive friction can further cause flexible instruments to buckle (Fig. 4.1b): the shaft accumulates elastic energy until suddenly released, potentially exerting intense forces that can cause catastrophic luminal damage [22, 23] and, in robot-assisted procedures, further complicate instrument control [24, 25]. In catheterization specifically, friction can damage vessel walls, promoting post-operative thrombus formation, and dislodge calcified particles that can subsequently obstruct critical blood vessels (Fig. 4.1b).

However, while high friction is detrimental to the navigation of transluminal instruments, it becomes an asset when executing surgical operations such as cutting, puncturing, or suturing. All these operations require frictional anchoring between the instrument and the tissue in order to maintain position and apply force with precision [26–28]. In low-friction conditions, particularly within open cavities such as the heart, the instability of the instrument may hinder effective manipulation and compromise procedural success (Fig. 4.1c). This contradiction, where low friction is needed during navigation and high friction is needed when anchoring, is a significant



**Figure 4.1. Friction-induced complications in transluminal procedures and our proposed active lubrication approach.** (a) Catheter navigation through the femoral artery, with the inset illustrating asperity contact at the catheter–vessel interface. (b) Excessive friction can cause vessel injury, catheter buckling, and dislodgement of calcified deposits. (c) Insufficient friction reduces instrument stability and precision. (d) The active lubrication concept integrates discrete friction-control modules that generate ultrasonic vibrations, creating a pressurized fluid film at the interface for frictionless navigation. Deactivating the vibrations restores frictional contact, anchoring the catheter at the target site. Friction state is controlled simply by toggling the vibration on and off.

ant design challenge for instruments, which must strike a balance between ease of navigation and stability.

Current practices rely on passive lubrication strategies to mitigate the risks of friction in transluminal interventions. To reduce friction during navigation, conventional strategies use liquid lubricants [29–33] or self-lubricating coatings, such as hydrophilic coatings that become slippery when exposed to a wet surface [34, 35]. However, these methods have notable limitations. Add-on lubricants are often short-lived, gradually losing effectiveness as they are wiped away during navigation [29]. Hydrophilic coatings, though effective in specific procedures such as urology, lack versatil-

ity for broader transluminal applications. Crucially, neither approach offers dynamic control over friction—they cannot switch between low-friction and high-friction states based on the procedural phase. The inability of these techniques to meet the changing frictional demands of transluminal interventions highlights the need for active lubrication strategies that enable real-time, on-demand friction control for safer navigation and more stable procedure execution.

This chapter introduces an active lubrication concept for transluminal instruments that enables real-time control of friction via ultrasonic lubrication. In this concept, we use high-frequency (ultrasonic) surface vibrations to create a pressurized squeeze film of fluid at the interface between the instrument and tissue, which actively lubricates the interface on demand, as illustrated in Fig. 4.1(d). By dynamically switching friction states, the instrument can navigate safely to its destination and then transition to a stable configuration for surgical tasks. We present a versatile active lubrication sheath that modulates friction along the instrument shaft on demand, building upon our initial demonstration [36]. We show how effective active lubrication modules are at changing friction under varying mechanical loads, surface conditions, and anatomical curvatures. Finally, we validate the system in a cardiac catheterization use case using *ex vivo* porcine aorta tissue, and assess its thermal safety during contact with biological tissues.

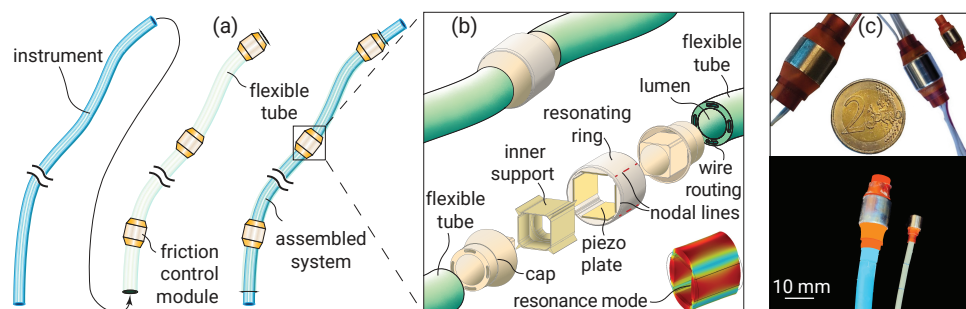
The remainder of this chapter is organized as follows. Section 4.2 introduces the working principle and design of the active lubrication sheath. Section 4.3 describes the experimental methods used to evaluate its performance. Section 4.4 presents the results and discusses their implications, and Section 4.5 concludes the chapter.

## 4.2. Working Principle and Design of Active Lubrication Sheath

### 4.2.1. Ultrasonic Lubrication

When two surfaces are pressed together, contact occurs only at a finite number of microscopic roughness asperities. These asperities in contact form micro-junctions, which collectively constitute the real area of contact that carries the applied load. These micro-junctions govern sliding friction, as the deformation and shearing of these micro-junctions when relative sliding occurs dictates the friction force experienced by the two objects in contact. The remaining regions of the interface, where microscopic asperities are not in contact, remain separated by small gaps, where fluid from the surrounding medium, whether air or liquid, is trapped.





**Figure 4.2. Design concept and prototypes.** (a) The proposed active lubrication sheath featuring discrete friction control modules along its shaft. The instrument inserted into this sheath can be actively lubricated on demand. (b) Design of the friction control modules and resonance mode of the resonating ring. (c) Top: 15 mm, 10 mm, and 3 mm prototypes of the friction control modules. Bottom: fully assembled sheath with friction control module at the tip, 10 mm and 3 mm size.

Ultrasonic lubrication exploits this trapped interfacial fluid. When transverse ultrasonic vibrations are applied to one of the contacting surfaces, the trapped fluid is repeatedly compressed and released. Consequently, the oscillatory motion generates a net positive pressure due to the non-linear response of the fluid to the periodic compression. This positive net pressure creates a fluid cushion between the two surfaces that partially levitates the vibrating surface away from the counter surface, decreasing asperity contact and thus reducing friction during sliding [37, 38].

The resulting friction reduction increases monotonically with vibration amplitude [38], allowing friction to be controlled by adjusting the excitation amplitude. With the vibrations off, the interface remains in a high-friction contact state. When the vibrations are activated, the squeeze film supports part of the normal load and shifts the interface toward a low-friction state. This principle, commonly referred to as ultrasonic friction modulation or ultrasonic lubrication, has been applied in tactile displays [39], contactless manipulation [40], and squeeze-film bearings [41, 42]. We recently demonstrated that it also remains effective in liquid environments [43], making it particularly relevant to transluminal instruments operating against hydrated biological tissue.

#### 4.2.2. Conceptual Design of the Active Lubrication Sheath

The active lubrication sheath features discrete friction control modules distributed along its length, enabling localized, on-demand friction control. When switched on,

these modules generate controlled ultrasonic vibrations that induce fluid film lubrication at the module-tissue interface, which transforms the instrument into a low-friction mode for frictionless navigation, as illustrated in Fig. 4.1(d). Once switched off, the modules re-establish contact with the luminal wall, creating friction-based anchoring for stable execution, as illustrated in Fig. 4.1(d).

The sheath can function as a catheter or be used to transform existing instruments into actively lubricated tools by inserting them into the sheath, as illustrated in Fig. 4.2(a). The modules are slightly larger in diameter than the sheath, to ensure that contact primarily occurs between the modules and the luminal walls, minimizing direct contact between the sheath body and tissue. Additionally, the sheath body is coated with an ultra-low friction material similar to [23] to ensure smooth interaction with the luminal wall, if contact occurs.

To effectively generate the desired active ultrasonic lubrication effect, the friction control modules should achieve vibration amplitudes of  $\geq 2 \mu\text{m}$  at an ultrasonic frequency  $\geq 20 \text{ kHz}$  [44, 45].

### 4.2.3. Detailed Design and Simulation

Each friction control module generates ultrasonic transverse vibrations through a resonating ring structure equipped with four piezoelectric plates, as shown in Fig. 4.2(b). When these plates are excited at the resonant frequency of the structure, the latter vibrates along its second flexural resonance mode, as depicted in Fig. 4.2(b). The resonance frequency is determined by the diameter of the ring and wall thickness. The resonating ring is supported by a separable inner support structure made from a self-lubricating material, featuring needle-like edges that hold the ring at the nodal lines of its vibration mode, where the vibration amplitude is near zero. These needle-like supports give way to the nodal lines to rotate freely which minimises the interference with the vibration. Additionally, the separable support facilitates the assembly of the piezoelectric plates, a crucial feature when miniaturizing the modules. To maintain an unobstructed instrument lumen, we incorporated specific wire-routing channels into the side cap and sheath wall [46], allowing the power wires to be routed through them.

Three ring sizes, with diameters ranging from 3 mm to 15 mm, were designed to operate in the second flexural resonance mode at a frequency above 20 kHz while generating vibration amplitudes exceeding  $2 \mu\text{m}$ . Finite element analysis (FEA) guided

the detailed ring design and provided estimates of its resonance characteristics. We first performed an eigenfrequency analysis to identify the resonance associated with the target second flexural mode. A subsequent frequency-response analysis evaluated the vibration amplitude and deformation pattern under harmonic excitation. By iteratively adjusting the ring dimensions, we obtained designs that met the frequency and amplitude requirements for ultrasonic lubrication. The 10 mm ring was selected for detailed simulation and subsequent experimental characterization, as it provides a practical balance between relevance to transluminal instruments and ease of fabrication and experimentation. For this design with titanium, the model predicted a second flexural resonance at 22.7 kHz and a peak vibration amplitude of approximately 4  $\mu\text{m}$  per 100 V.

#### 4.2.4. Prototype Development

We implemented the designed ring structures in titanium using wire electrical discharge machining. Four 1 W ceramic piezoelectric plates (SMPL60W5T03R112, STEMiNC, Davenport, IA, USA), each 0.3 mm thick, were then cut to size and manually bonded around each ring using a custom wedge fixture that applied uniform pressure during curing. After bonding, 0.5 mm power wires were soldered to the plates and routed through the internal channels and side caps. Following this process, we fabricated three prototypes with diameters ranging from 3 mm to 15 mm to demonstrate the scalability of the design, as shown in Fig. 4.2(c). The 10 mm prototype was used for the subsequent characterization and validation experiments.

### 4.3. Experimental Evaluation

We experimentally evaluated the active lubrication system through three sets of experiments. First, we characterized and validated the vibration response of the fabricated modules. Second, we evaluated their active lubrication performance through a series of sliding friction experiments under varying loading, lubricity, and curvature conditions, including validation on *ex vivo* tissue. Third, we assessed the thermal safety of the modules via thermal imaging experiments. Additionally, we conducted a catheter insertion experiment to demonstrate the practical effect of active lubrication on buckling.

#### 4.3.1. Vibration Characterization

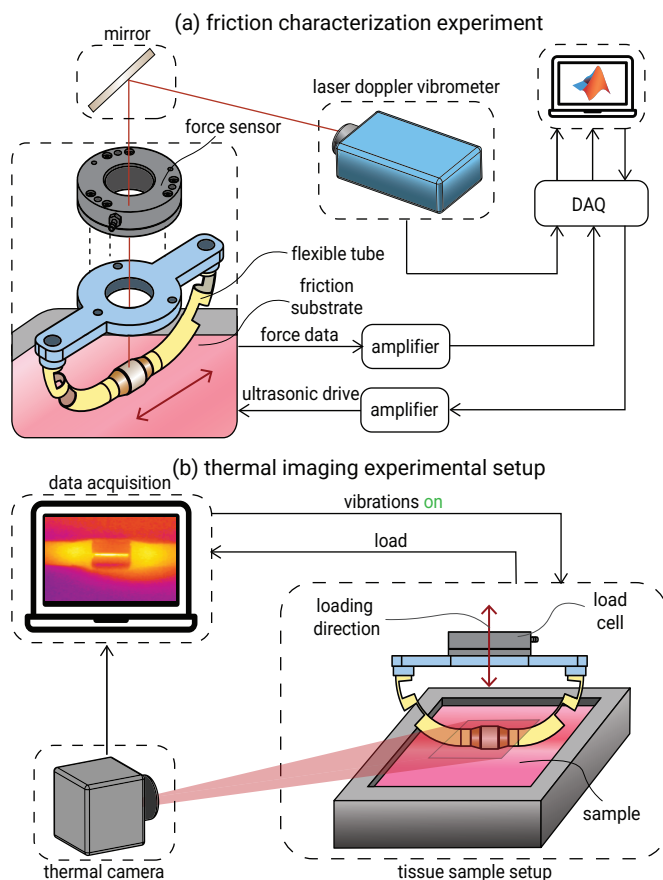
We characterized the vibration response of the fabricated modules using a single-point laser Doppler vibrometer (LDV) (OFV5000, Polytec GmbH, Germany). To access the full ring surface, the module was mounted on a servo motor that rotated it around its circumference, while a dual-axis galvanometer scanner (ScannerMAX Saturn 5B, Edmund Optics Inc., NJ, USA) steered the LDV beam along its length, following an approach similar to [36]. A MATLAB interface synchronized the servo motor, scanner, LDV, and excitation signal through a National Instruments data acquisition card (USB-6356, National Instruments, TX, USA).

The characterization followed three steps. First, frequency sweeps at the centers of the four anti-nodal lines identified the resonant frequency. Second, a voltage sweep at resonance established the relationship between the input voltage and peak vibration amplitude. Third, a full-surface scan reconstructed the vibration mode and quantified the effective vibrating area. During this scan, the ring rotated in  $5^\circ$  increments while the LDV recorded the vibration response at ten points along each longitudinal scan line. The effective vibrating area was then calculated as the portion of the ring surface vibrating at or above the target amplitude of  $2\text{ }\mu\text{m}$ .

#### 4.3.2. Active Lubrication Characterization

We characterized the active lubrication performance using the sliding friction setup shown in Fig. 4.3(a). The resonating ring was mounted on a six-axis force sensor (Nano43, ATI Industrial Automation, NC, USA), and the ring-sensor assembly was attached to a manual linear slider that controlled the normal load. A motorized linear stage moved the opposing substrate against the ring surface, while the force sensor recorded the friction force and the LDV measured the vibration amplitude from the top of the ring.

The characterization followed two test series. In both, the substrate traveled 15 mm at 1 mm/s under the prescribed normal load, and each condition was repeated six times. First, the input voltage was increased linearly from its minimum to maximum value over 5 s during sliding to establish the relationship between vibration amplitude and friction under different loading and surface lubricity conditions. Second, substrates with different curvatures were tested by activating the module at the maximum input voltage 5 s after sliding began. The resulting friction reduction was then



**Figure 4.3. Experimental setups.** (a) Friction characterization setup combining force and vibration measurements during substrate sliding. (b) Thermal imaging setup for assessing thermal safety of ultrasonic lubrication.

normalized by the corresponding vibration amplitude to quantify the friction reduction capacity (reduction percentage per unit amplitude).

### 4.3.3. *ex vivo* Tissue Validation

We validated the friction-control modules on hydrated *ex vivo* biological tissue using porcine aorta samples. Aortas from approximately six-month-old pigs were obtained from a local abattoir within 2 h of slaughter and transported in phosphate-buffered saline (PBS) at 4°C. After residual tissue was gently removed, rectangular sections approximately 10 mm in size were cut along the arterial axis and stored at -80°C for

no longer than 30 days. Before testing, the samples were fully thawed and rehydrated in PBS.

Four tissue samples were tested under a normal load of 1 N. Each sample was cut open, flattened, and kept hydrated with PBS throughout the experiment, following [11, 47]. Six sliding trials were performed along each path. The first three trials were used to condition the tissue, while the final three were included in the analysis. Three distinct paths were tested on each sample to assess path dependence. During each trial, friction was first measured with the vibrations inactive and then immediately after activating the module, enabling a direct within-trial comparison of the coefficient of friction between the inactive and active states.

#### 4.3.4. Thermal Safety Assessment

To assess thermal safety, heat generation at the module–tissue interface was monitored using a thermal camera (thermoIMAGER TIM 160S, Micro-Epsilon, NC, USA), as shown in Fig. 4.3(b). The tissue was hydrated with PBS, and the module was loaded against its surface at 1 N. The module was then activated without sliding while the interface temperature was continuously recorded at an ambient temperature of 20–22°C. Six tissue samples were tested, with three repetitions per sample.

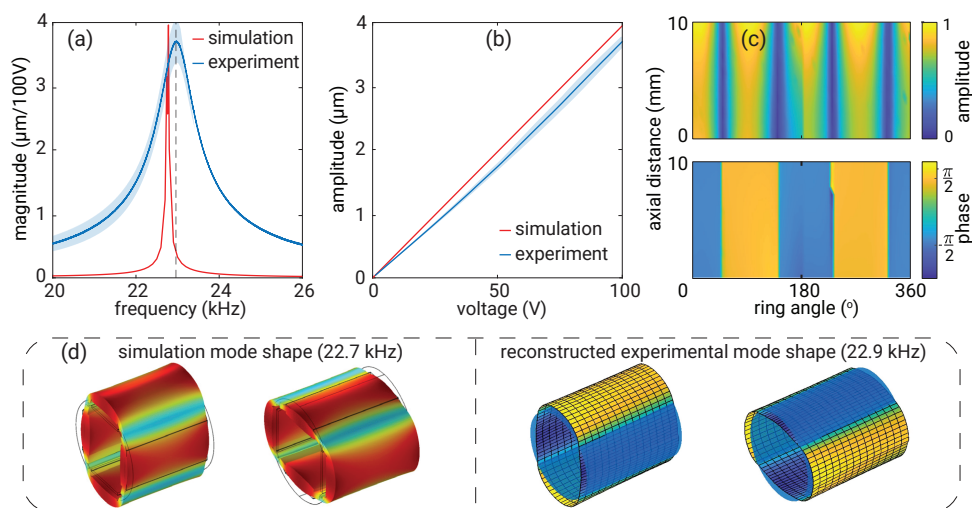
#### 4.3.5. Catheter Buckling Demonstration

Finally, we integrated the active lubrication module at the tip of a catheter to qualitatively demonstrate its effect on buckling during insertion into a 3D-printed lumen model. The catheter base was deliberately misaligned with the lumen axis to increase friction at the tip and create a challenging insertion condition. We compared insertions with and without activation to demonstrate how reducing friction at the catheter tip influences buckling and ease of insertion.

### 4.4. Results and Discussion

#### 4.4.1. Vibration Characterization

The resonant frequency of the 10 mm prototype was measured at 22.9 kHz, with a linear vibration amplitude of slope  $3.7 \mu\text{m}/100 \text{ V}$ , in close agreement with the FEA predictions of 22.7 kHz and  $\approx 4 \mu\text{m}/100 \text{ V}$  (Fig. 4.4(a)(b)). Surface scanning confirmed

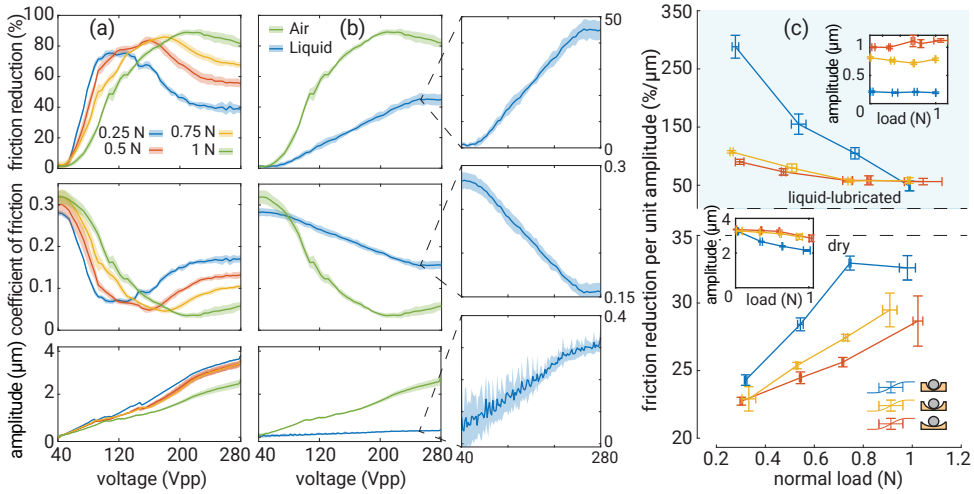


**Figure 4.4. Vibration characterization of the active lubrication modules.** (a) Frequency response of the resonating ring showing agreement between simulation and experiment. (b) The relationship between amplitude of vibration and input voltage applied to the piezoelectric actuators for simulated and experimental results (c) Mode shape of the resonating ring is the second flexural resonance mode matching simulation (d) comparison between the predicted mode shape by simulation and the experimentally reconstructed mode shape of the resonating ring prototype. (See Supplementary Video 1 for the video demonstration.)

that the excited mode is the second flexural resonance mode (Fig. 4.4(c)(d)), with  $\approx 84\%$  of the ring surface vibrating at or above the  $2 \mu\text{m}$  threshold required for effective ultrasonic lubrication. The minor amplitude discrepancy between simulation and experiment ( $3.7$  vs  $4 \mu\text{m}/100 \text{ V}$ ) can be attributed to fabrication tolerances and differences in material properties between model and prototype.

#### 4.4.2. Friction Modulation Characterization

The friction modulation experiment in dry contact revealed a linear correlation between the amplitude and the input voltage (Fig. 4.5(a)), consistent with the contact-free response in Fig. 4.4(b), albeit with a shallower slope. As the load increases, the slope of the correlation shifts downwards due to the greater damping associated with higher loads. The corresponding friction reduction was robust to load, consistently achieving a similar maximum friction reduction in all loading conditions  $\approx 80 - 82\%$ , matching the maximum friction reduction achieved by ultrasonic haptic touchscreens [39, 44]. However, increasing the normal load causes a horizontal offset in the friction reduction curve towards higher voltages. This offset is due to the necessity for higher vibration amplitudes to counteract the extra load,



**Figure 4.5. Active lubrication performance under varying contact conditions.** (a) Friction reduction, coefficient of friction, and vibration amplitude as functions of input voltage at four normal loads. The vibration amplitude increases with voltage but decreases with increasing load, while friction reduction rises to a maximum before slightly declining at higher amplitudes. (b) Comparison between air and liquid environments, showing a more linear response and greater friction reduction for a given vibration amplitude in liquid. (c) Friction-reduction capacity across different normal loads and surface curvatures under dry and liquid-lubricated conditions. Under dry contact, the capacity increases with curvature and load, whereas under liquid lubrication it remains higher overall but decreases at larger loads. The insets show the corresponding vibration amplitudes. All conditions were repeated six times; lines and points represent the mean, while shaded regions and error bars indicate the standard deviation.

to maintain similar levels of friction reduction. Beyond the peak, friction reduction consistently declined across all loads, suggesting a saturation of the load-carrying capacity of the air squeeze film. Once the film pressure saturates, further energy input is converted to heat, increasing air viscosity, and consequently dissipation, which reduces the load-carrying capacity of the film and increases friction.

In liquid-lubricated contact, the amplitude-to-voltage mapping followed a consistently linear response but with an amplitude of  $0.3 \mu\text{m}$  at maximum input voltage — approximately seven times lower than in air (Fig. 4.5(b)). Friction reduction was negligible below  $\approx 80 \text{ V}$ , below which amplitude was insufficient to overcome interfacial capillary adhesion, then followed a linear trend before saturating at  $\approx 42\%$ . While this maximum is roughly half the dry-contact value, the friction reduction per unit amplitude was significantly higher in liquid — consistent with the superior load-carrying capacity of incompressible liquid squeeze films predicted by theory [43]. We also observe that liquid-lubricated contact exhibits a linear behavior for the full range of input voltage in comparison to dry contact. Importantly, in the liquid case, saturation



of the reduction of friction is caused by a saturation of the vibration amplitude, unlike in the dry condition where the saturation comes from the load carrying capacity of the air film.

#### 4.4.3. Effect of Surface Curvature

In dry contact conditions, we observed that the vibration amplitude decreases with increasing load in all three curvature profiles, as shown in the inset of Fig. 4.5(c). This can be attributed to an increase in viscous losses caused by higher loads. Additionally, at a given load, the vibration amplitude decreases with increasing diameter ratio, which can be attributed to the larger contact area and the resulting increase in damping. For the friction reduction capacity, we found that at the same load, surfaces with higher curvature exhibited greater friction reduction capacity, which can be explained by the greater contact area generating a proportional increase in the squeeze film force [43], as shown in Fig. 4.5(c). Moreover, the friction reduction capacity improved with increasing load across all curvature profiles. This trend is likely due to the reduced interfacial gap thickness between the two surfaces at higher loads, enhancing the squeeze film force, which results in greater friction reduction at the same vibration amplitude.

Looking at the liquid-lubricated contact in Fig. 4.5(c), we notice that the vibration amplitude, shown in the inset, remains almost constant across all loads. This constancy is likely due to the vibrating module being levitated and vibrating on top of a liquid squeeze film. In addition, the friction reduction capacity is reduced as the load increases, most likely because at higher loads, the interfacial gap thickness between the two surfaces becomes smaller, which in turn impedes the liquid from flowing in and out of the interface. This free movement of the liquid is a crucial factor for squeeze film pressure to develop in liquid films [43]. We observe that the most curved interface in the liquid-lubricated case initially exhibits significantly larger friction reduction capacity as compared to the other two profiles. It can be explained by the larger surface area contributing to an increased squeeze film force. However, the friction reduction capacity rapidly declines as the load increases due to the higher resistance for the liquid to move across the film. This larger resistance causes the liquid to lose more energy and thus reduces its load-carrying capacity.

Based on the linear correlation, between friction and the vibration amplitude, we observed in the friction modulation experiment Fig. 4.5(b), one can assume that the

behavior in the liquid-lubricated case, shown in Fig. 4.5(c) holds across different input voltages. However, for dry contact, the same conclusion may only be applied in the linear region of the response in Fig. 4.5(a), while the other region is likely voltage-dependent.

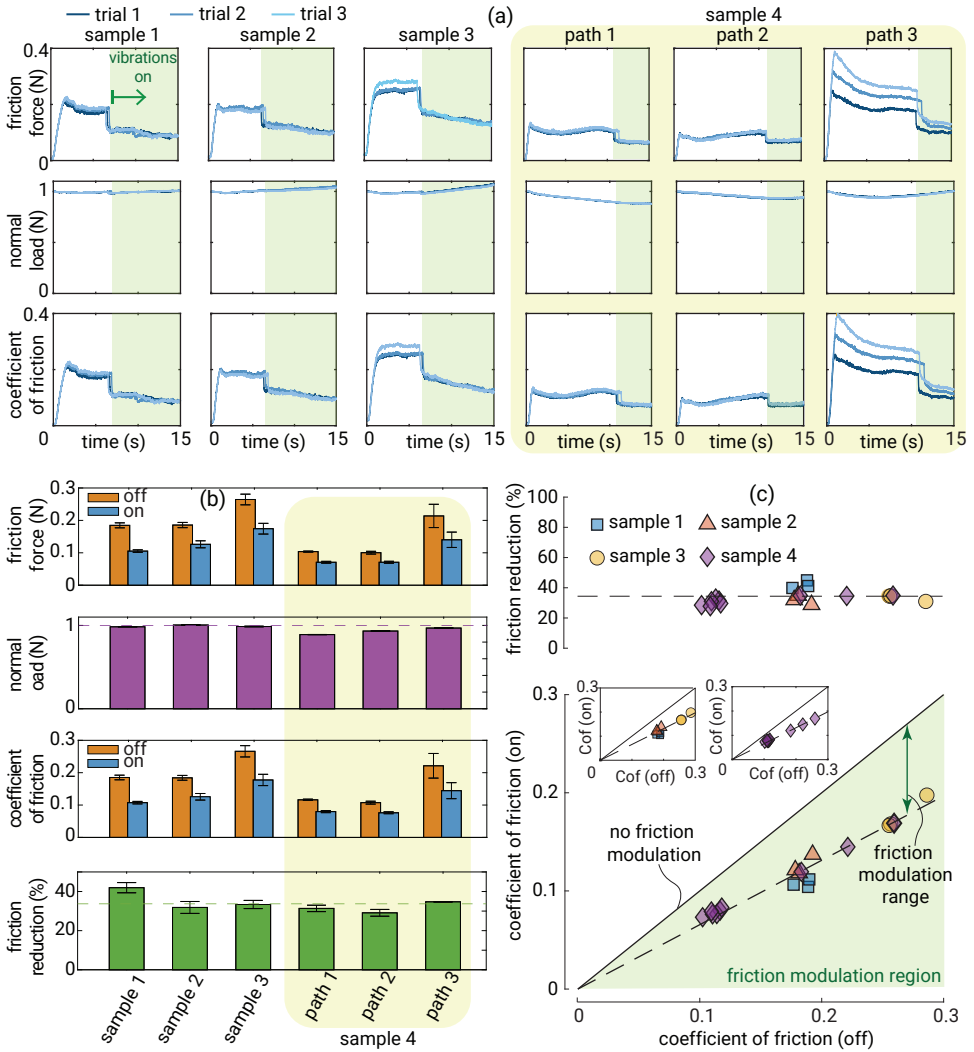
We conclude that highly curved surfaces are more favorable for friction reduction in both dry and liquid-lubricated environments. This conclusion is supported by the fact that friction force does not depend on the apparent area of contact. As curvature increases, the apparent area of contact increases, enhancing the squeeze-film force without affecting the friction force, resulting in a net increase in the friction reduction capacity.

#### 4.4.4. *ex vivo* Tissue Validation

Upon enabling vibration, the coefficient of friction dropped immediately to a lower level as shown in Fig. 4.6(a), achieving friction reduction up to 42% and an average of 38% across all tissue samples and paths at maximum vibration amplitude as shown in Fig. 4.6(b).

Looking closely at the results in Fig. 4.6(a)&(b), we first observe that normal load varies over time across the different samples and paths with small variance across trials. These variations result from the non-uniform topography of the tissue samples along the sliding path. Second, we observe that the starting coefficient of friction (before vibrations are activated)  $\mu_{\text{off}}$  varied between samples, paths and between trials (example in Fig. 4.6(a) - Sample 4-Path 3). This variation across samples and paths can be attributed to the uneven surface topography of tissue samples which resulted in uneven distribution of the surface lubrication layer. In addition, we observed that the tissue samples were drying out over time, resulting in increasing friction levels across the different trials.

This seemingly undesired variance in the coefficient of friction revealed an interesting correlation between the coefficient of friction in the off condition (before vibrations are activated)  $\mu_{\text{off}}$  and the coefficient of friction in the on condition (after vibrations are activated)  $\mu_{\text{on}}$  as shown in Fig. 4.6(c). The coefficient of friction in the on condition  $\mu_{\text{on}}$  is positively correlated to the coefficient of friction in the off condition  $\mu_{\text{off}}$  and this correlation holds for both different samples and different paths as shown in the insets of Fig. 4.6(c). This correlation is well explained by the model of Wierblewski et al. [38] in which they showed using multi-scale contact theory [48–50] and



**Figure 4.6. Friction validation on ex vivo porcine aorta tissue.** Each experiment was repeated six times; the first three repetitions conditioned the tissue and the last three were used for analysis. (a) Recorded friction force, normal load, and coefficient of friction across different samples and paths. Upon activating vibrations, friction drops immediately, demonstrating the active lubrication effect. (b) Comparison across samples and paths (mean  $\pm$  standard deviation). Variability in friction and normal load is attributed to non-uniform tissue topography and partial drying of the samples over time. (c) Linear correlation between the coefficient of friction in the off condition  $\mu_{\text{off}}$  and the on condition  $\mu_{\text{on}}$ .

the squeeze film theory [37, 51] that the coefficient of friction after enabling vibrations  $\mu_{\text{on}}$  is proportional to the coefficient of friction without vibrations  $\mu_{\text{off}}$  as follows:

$$\mu_{\text{on}} = \beta \cdot \mu_{\text{off}} \cdot \exp\left(\frac{\alpha}{N}\right) \quad (4.1)$$

where  $\alpha$  is the vibration amplitude,  $N$  is the normal load and  $\beta$  is a proportionality constant that depends on surface roughness and material elasticity. This equation implies that the slope of the friction modulation line depends on the vibration amplitude and normal load. At identical vibration amplitude and load, the range between  $\mu_{\text{off}}$  and  $\mu_{\text{on}}$ , as shown in Fig. 4.6(c), increases linearly with the increase in  $\mu_{\text{off}}$ , while the friction reduction percentage (%) remains constant.

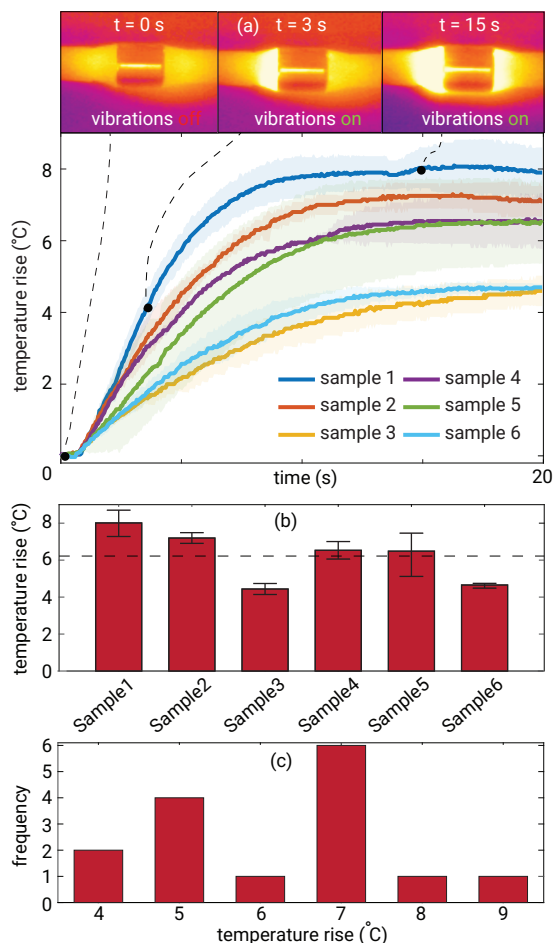
This correlation between the coefficient of friction in on ( $\mu_{\text{on}}$ ) and off ( $\mu_{\text{off}}$ ) conditions, shown in Fig. 4.6(c), provides key design insights for optimizing the active lubrication system. Specifically, it is beneficial to maximize both the starting coefficient of friction ( $\mu_{\text{off}}$ ) and the vibration amplitude. Higher  $\mu_{\text{off}}$  increases the difference between  $\mu_{\text{off}}$  and  $\mu_{\text{on}}$  at maximum vibration amplitude, enabling more effective switching between friction states. Similarly, maximizing vibration amplitude increases the controllable range between  $\mu_{\text{off}}$  and  $\mu_{\text{on}}$  for a given starting coefficient of friction  $\mu_{\text{off}}$ . Combining both factors can potentially result in a wider controllable range, with higher friction in the "off" state and lower friction in the "on" state. However, further investigation is needed to determine the optimal controllable range and the acceptable limits for interacting with tissues. That is to find the most effective range of controllable friction states ( $\mu_{\text{off}}$  and  $\mu_{\text{on}}$ ) that minimises tissue damage while ensuring stability during catheterization.

#### 4.4.5. Thermal Safety

Upon activation, the module–tissue interface temperature rose immediately and reached a steady state after approximately ten seconds, as shown in Fig. 4.7(a). We found the average temperature rise to be  $\approx 6.2^\circ\text{C}$ , as shown in Fig. 4.7(b), with  $7^\circ\text{C}$  being the most frequent temperature rise as shown in Fig. 4.7(c). This temperature rise translates to  $\approx 44^\circ\text{C}$  inside the human body which is within the thermal safety range for the human body [52, 53], suggesting the potential thermal safety of this technology for transluminal surgery applications.

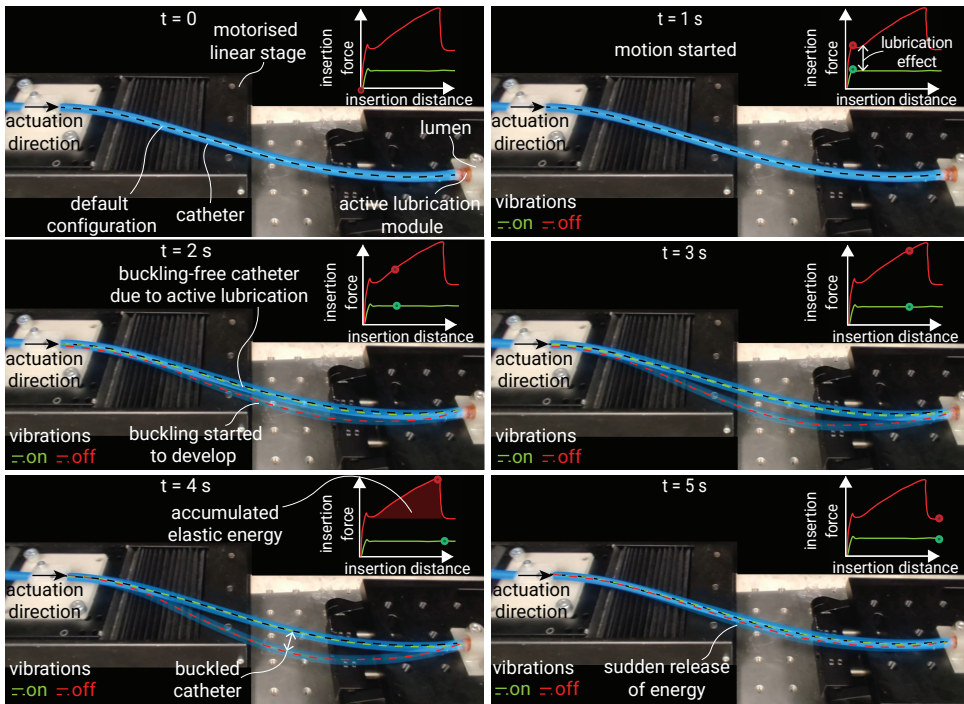
#### 4.4.6. Catheter Buckling Demonstration

As shown in Fig. 4.8, without lubrication, the catheter tip experienced higher friction, causing the catheter to buckle. This buckling resulted in elastic energy accumulation in the shaft of the catheter until it was suddenly released; rapidly advancing the catheter tip forward. This sudden release of energy can exert excessive forces on the



**Figure 4.7. Thermal safety assessment results.** Each experiment was repeated three times; solid lines and bars represent the mean, shadings and error bars the standard deviation. (a) Time response of the temperature rise across six tissue samples with representative thermal images. Temperature increases upon activation and reaches steady state within approximately 10 s. (b) Steady-state temperature rise per sample, averaging  $6.2^{\circ}\text{C}$ . (c) Histogram of temperature rise, with  $7^{\circ}\text{C}$  being the most frequent value, within the established thermal safety limit [52, 53]. (See Supplementary Video 2.)

luminal wall, potentially causing catastrophic lumen dissection or rupture [22, 23]. In contrast, with the lubrication module activated, the catheter tip moved smoothly through the lumen without buckling, as seen in Fig. 4.8, maintaining low frictional interaction with the lumen wall. This demonstration underscores the positive impact of the active lubrication system in reducing catheter buckling, potentially enhancing insertion and navigation, and thus ultimately improving the safety and efficacy of transluminal procedures.



**Figure 4.8. Demonstration of active lubrication preventing catheter buckling during insertion.** A catheter with a friction-control module at its tip was inserted into a 3D-printed lumen, with the base deliberately misaligned to promote frictional wall interaction. Without lubrication, the catheter buckled, accumulating elastic energy in the shaft until it was suddenly released, potentially exerting excessive shock loads on the lumen wall. With lubrication activated, the catheter navigated smoothly without buckling. (See Supplementary Video 3.)

#### 4.4.7. Limitations and Future Work

The current study has several limitations that should be considered when interpreting the results. Friction was characterized on rigid 3D-printed substrates and one soft tissue type, porcine aorta. Although these represent calcified and healthy vascular tissue, respectively, they do not cover the full range of tissue properties encountered in transluminal procedures. Thermal safety was also tested under static and dry conditions at room temperature, without sliding or liquid flow. The measured temperature rise may therefore differ from that at body temperature and likely overestimates the heating expected during in vivo use, where blood flow would remove heat through convection. In addition, the contact point between the module and the tissue would continuously shift during catheter navigation, limiting the duration of local heat exposure. These effects need to be examined under more physiologically real-

istic conditions. The current prototype also lacks waterproofing, which prevents fully submerged operation, while the spacing, diameter, and number of modules along the sheath have not yet been optimized. Finally, the effects of prolonged ultrasonic vibration on tissue have not yet been studied at the histological level.

Future work will extend the validation to additional tissue types, including intestinal tissue, to assess how well the approach transfers to endoscopic and other transluminal procedures. Histological analysis will also be used to determine whether prolonged exposure to the vibrations causes tissue damage. A waterproof module will be developed to support fully submerged testing, followed by in vivo animal studies using complete instrument assemblies. The spacing, diameter, and number of modules along the sheath will also be refined to improve friction control without reducing instrument flexibility. In parallel, thermal testing will be repeated at body temperature and under flow or perfusion to better represent physiological conditions and define a more reliable safety baseline.

The same principle may also be useful beyond cardiovascular applications in procedures where friction affects instrument navigation, such as flexible endoscopy and bronchoscopy. The characterization across normal load, lubrication, and surface curvature provides a basis for adapting the module design to these different anatomical and procedural conditions.

## 4.5. Conclusion

This chapter presented an active lubrication sheath for transluminal medical instruments that controls friction on demand through discrete ultrasonic modules distributed along the shaft. The modules operated at 22.9 kHz, reached a peak vibration amplitude of  $3.7 \mu\text{m}/100 \text{ V}$ , and provided effective lubrication over approximately 84% of their surface. Tribometric experiments showed friction reductions of up to 82% on rigid surfaces and 42% on PBS-lubricated ex vivo porcine aorta tissue, representing simplified calcified and healthy vascular interfaces, respectively. The results also showed a clear relation between ( $\mu_{\text{off}}$ ) and ( $\mu_{\text{on}}$ ), indicating that a wider controllable friction range requires both a higher initial coefficient of friction and a larger vibration amplitude. Active lubrication was most effective under liquid-lubricated conditions and generally improved with surface curvature, possibly because of the larger contact area. The average temperature rise was  $6.2^\circ\text{C}$  under static and dry test conditions, although physiological testing is still needed to assess the thermal response

during use. In the catheter insertion experiment, the reduced wall friction lowered the tendency to buckle and enabled smoother advancement through the lumen. These findings support the use of ultrasonic lubrication in cardiovascular instruments and other transluminal devices, while future work should focus on waterproof modules, physiological thermal testing, additional tissue types, histological assessment, and optimization of the module arrangement along the shaft.

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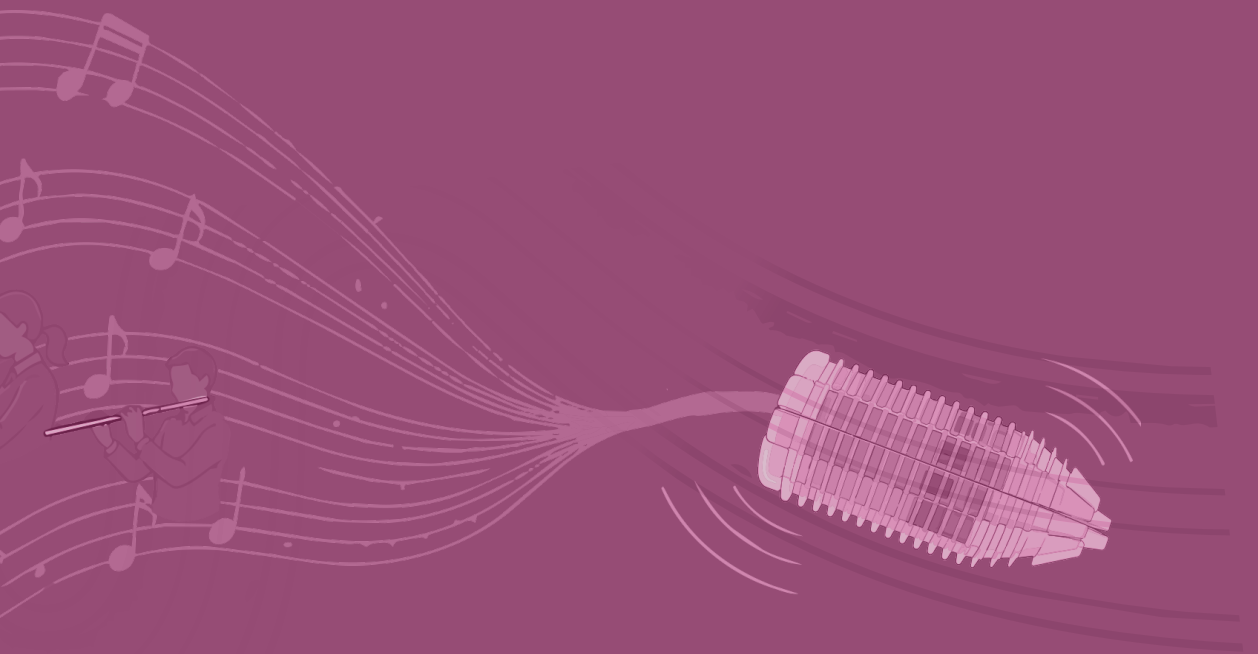
# PART II

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# Modulate Friction to Propel

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# CHAPTER 5

## Self-propelling Mechanisms for Colonoscopy: A Review

Under review as: M. A. Atalla, D. van Bommel, M. Wiertlewski, A. Sakes and P. Breedveld,  
*'Traversing the Colon: A Review of Self-propelling Mechanisms for Robotic Colonoscopy'*.



## Abstract

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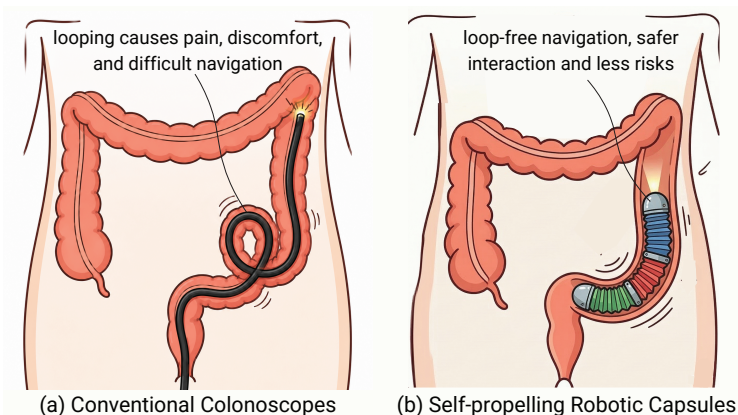
Self-propelling capsules aim to reduce patient pain and improve procedural outcomes in colonoscopy by replacing conventional push-driven insertion with propulsion generated at the device-lumen interface. These capsules use onboard self-propelling mechanisms to pull themselves forward through interaction with the colon wall, rather than transmitting force from the proximal end, eliminating shaft looping and reducing mechanical strain on the colon. However, the colon presents a particularly demanding locomotion environment, where lubrication, compliance, and large deformations make reliable propulsion difficult to achieve. This chapter systematically reviews the state-of-the-art self-propelling mechanisms used in tethered self-propelling capsules for colonoscopy, examining their fundamental locomotion principles and performance across different evaluation stages. Following a systematic search of the Scopus database, a total of 111 articles were included and classified according to their underlying propulsion strategy. The identified systems are first broadly divided into wall-mediated and non-wall-mediated approaches. Wall-mediated devices are further organized based on the presence of local tangential slip against the tissue; this encompasses mechanisms that operate without slip (achieved through intermittent anchoring or continuous rolling) and those that operate with slip (relying on actively changing friction or traveling waves). Conversely, non-wall-mediated systems utilize self-growing or fluid-driven mechanisms to navigate the complex anatomy without relying on propulsive wall traction. By organizing the field around the mechanics of propulsion and relating these locomotion principles to reported performance, this chapter establishes the design framework for Part II and identifies key opportunities for the development of friction-enabled, self-propelling colonoscopic robots.

## 5.1. Introduction

Colorectal cancer (CRC) remains a major global health burden, ranking among the most commonly diagnosed malignancies worldwide and one of the leading causes of cancer-related death [1, 2]. Timely screening is therefore essential, and colonoscopy remains the clinical gold standard because it uniquely enables both detection and removal of precancerous lesions within the same procedure [3, 4]. Owing to this dual diagnostic and therapeutic capability, colonoscopy plays a central role in reducing CRC incidence and mortality and is performed on a massive scale worldwide [5, 6]. Despite this clear clinical value, however, the procedure remains mechanically challenging and burdensome for both patients and clinicians.

Conventional colonoscopy relies on manually advancing a flexible endoscope through the tortuous, deformable, and highly compliant colon [7]. Because insertion is achieved by pushing from the proximal end, the shaft frequently bends and loops inside the bowel, particularly in mobile segments such as the sigmoid colon [8, 9], as illustrated in Fig.5.1(a). This looping stretches the colon wall, contributes substantially to patient pain and discomfort, and forces the operator into repeated cycles of pushing, pulling, and twisting to recover a path forward [8, 10]. In parallel, scope handling itself is ergonomically demanding, requiring simultaneous manipulation of insertion forces and control knobs while mentally mapping hand motions to distal tip motion, which increases the cognitive and physical load on the clinician [11–13]. Together, these mechanical and ergonomic limitations motivate the search for alternative colonoscopic technologies that can reduce pain, improve controllability, and ease procedural difficulty.

Robotic colonoscopy aims to address these limitations by relocating actuation from the proximal end to the distal portion of the device, such that navigation is no longer dominated by the external pushing of a long flexible shaft. By generating motion closer to the site of interaction, robotic systems aim to reduce loop formation, decrease global stretching of the colon, and improve control of the device inside the lumen. Within this broader landscape, tethered self-propelling colonoscopic robots seek to achieve these benefits through an onboard self-propelling mechanism that allows the robot to pull itself forward along the wall without depending on proximal actuation, as illustrated in Fig.5.1(b). This self-propulsion eliminates looping and greatly reduces colon wall stretching, making these devices a promising option for less painful and more patient-friendly colonoscopy while preserving tethered functionality for therapeutic interventions [14, 15].



**Figure 5.1. Conventional colonoscopes versus self-propelling capsule robots for colonoscopy.** (a) Conventional colonoscopes are externally pushed to navigate the colon resulting in looping that stretch the colon wall and cause pain and discomfort. (b) Self-propelling robots pull itself along the wall of the colon resulting in loop-free navigation and consequently minimal pain and discomfort.

At a high level, self-propelling colonoscopic robots are designed to advance by mechanically interacting with the colon wall and using that interaction to pull themselves forward through the lumen. Rather than transmitting insertion force through a long shaft from outside the body, these devices incorporate an onboard propulsion mechanism that generates traction locally at the distal end while towing the body of the device and, in many concepts, an attached tether. In this architecture, the propulsion mechanism is not merely an actuation component but a defining functional element: it determines how traction is generated, how forces are transmitted to the tissue, how effectively the device can tow its tether, and how propulsion performance is balanced against safety, compactness, and clinical practicality. As a result, the future viability of self-propelling colonoscopy depends strongly on the underlying mechanism by which propulsion is achieved.

Several reviews have already contributed important perspectives to the field of robotic colonoscopy. Early work reviewed locomotion concepts for robotic colonoscopy at a time when the design space was still relatively limited [16]. More recent reviews have broadened the discussion toward comprehensive overviews of robotic colonoscopy technologies, device readiness, clinical alternatives, robotic gastrointestinal endoscopy in general, or specific subsets such as bioinspired and capsule-based systems [14, 17–23]. However, the field still lacks a review focused specifically on the self-propelling mechanisms for colonoscopy and their working principle. Such a perspective is important because, in self-propelling colonoscopic

robots, the propulsion mechanism is not merely one component among others, but the core element that determines how traction is generated, how the device interacts mechanically with the colon wall, how effectively it can tow a tether or a load, and how propulsion performance is balanced against safety, compactness, and clinical practicality. In addition, comparison across the literature is complicated by substantial differences in experimental validation methods, test environments, and reported metrics, making it difficult to determine whether differences in outcome truly reflect differences in mechanism performance.

Accordingly, this review aims to investigate self-propelling mechanisms for colonoscopy from the perspective of both working principle and experimental evidence. Specifically, it aims to identify the self-propelling mechanisms proposed in the literature and organize them into a meaningful taxonomy, to analyze the main advantages, limitations, and trade-offs associated with their different working principles, and to synthesize the reported experimental outcomes in order to assess which approaches appear more promising for the future of colonoscopy. In parallel, this review aims to examine how differences in validation techniques influence the interpretation and comparison of the reported results, providing a critical and contextualized perspective on the future of self-propelling mechanisms for colonoscopy.

The remainder of this chapter is organized as follows. We first describe the search methodology and inclusion criteria used to identify the relevant literature. We then present the classification framework adopted for self-propelling mechanisms for colonoscopy, followed by an in-depth review of each locomotion category, in which we examine how the different mechanisms were implemented and critically analyze their respective advantages and disadvantages. Next, we review the evaluation methods used across the literature in order to highlight the variability in experimental validation approaches. Building on this, the discussion section contextualizes the quantitative data reported in the reviewed studies by relating the performance of each mechanism category to the corresponding evaluation method, and by identifying the main pitfalls, limitations, and future opportunities in the field. Finally, the chapter closes with concluding remarks and directions for future work.

## 5.2. Search Methodology

We performed a search in the Scopus database to get a complete overview of the currently available literature on self-propelling mechanisms for robotic colonoscopy.

The criteria we used in our search included the locomotion capability represented by the terms (propel\* OR propul\* OR crawl\* OR locom\*), the type of device (robot\* OR capsule\* OR endosco\* OR colonosc\*), and the location in which these robots are utilized, focusing on colon-relevant terms (colon\* OR intestin\* OR gastrointestin\* OR bowel). The terms 'wireless' and 'WCE' (Wireless Capsule Endoscopy) were excluded to filter out passive, pill-cam style devices. Additionally, the terms 'micro\*' and 'magnetic' were excluded as we are not focusing on micro-scale robots or devices driven by external magnetic fields. The complete set of search terms is shown in Table 5.1. The search was limited to journal articles and conference papers published in English before the year 2026. The search was executed in the title, abstract, and keywords of Scopus. The following exact search query was used:

```
TITLE-ABS-KEY ( ( propel* OR propul* OR crawl* OR locom* ) AND ( robot* OR capsule* OR endosco* OR colonosc* ) AND ( colon* OR intestin* OR gastrointestin* OR bowel ) AND NOT ( wireless ) AND NOT ( 'WCE' ) AND NOT ( micro* ) AND NOT ( 'magnetic' ) ) AND PUBYEAR < 2026 AND ( LIMIT-TO ( DOCTYPE , "ar" ) OR LIMIT-TO ( DOCTYPE , "cp" ) ) AND ( LIMIT-TO ( LANGUAGE , "English" ) )
```

### 5.2.1. Inclusion and Exclusion Criteria

Articles were included if they detailed the design, development, or evaluation of a mechanism that facilitates self-propulsion in a robotic colonoscope. In this context, self-propulsion is defined as the device's ability to advance through the colon actively, without relying on an external pushing force (such as the manual insertion of a conventional rigid shaft). A colonoscopic robot is defined as a system capable of executing locomotion or navigational tasks autonomously or semi-autonomously. Furthermore, only full-text articles that were accessible and written in English were considered for inclusion.

### 5.2.2. Search Results and Article Selection

Our systematic search of the Scopus database initially yielded 372 articles. First, duplicate records were identified and removed. Next, the titles and abstracts of the remaining papers were screened for relevance based on our inclusion criteria. During this phase, review articles were also excluded, though their reference lists were cross-checked to ensure all relevant primary devices were already captured in our search. Finally, a rigorous full-text evaluation was conducted on the remaining candidates.

| Function | Device    | Location        | Exclusion |
|----------|-----------|-----------------|-----------|
| propel*  | robot*    | colon*          | magnetic  |
| propul*  | capsule*  | intestin*       | micro*    |
| crawl*   | endosco*  | gastrointestin* | wireless  |
| locom*   | colonosc* | bowel           | WCE       |

**Table 5.1.** Search terms used for the literature search.

After excluding studies that lacked sufficient mechanical detail, utilized passive locomotion, or did not meet the strict definition of self-propulsion, we identified 111 articles relevant for the purpose of this comprehensive review.

### 5.3. Classification of Self-Propelling Mechanisms

The self-propelling mechanisms of robotic colonoscopes can be broken down into two main components: the *thrust-generation mechanisms* and the *thrust-enhancement features*. The thrust-generation mechanism is the core physical action that actively drives the device forward through the colon. In contrast, thrust-aid features do not generate forward momentum on their own. Instead, they act as supporting elements that assist, bias, or stabilize the primary thrust. Making this distinction allows us to clearly separate a robot's main source of propulsion from the secondary design tweaks that improve its control and efficiency.

This review first categorizes systems based on their primary thrust-generation principle. At the highest level, locomotion is either *wall-mediated* (relying on traction against the colon wall) or *non-wall-mediated* (advancing without relying on wall traction). Within the *wall-mediated* category, mechanisms are divided by how they physically interact with the tissue. Systems that grip the tissue *without local tangential slip* primarily rely on *intermittent anchoring* (where discrete parts of the robot sequentially grab and release the wall) or *rolling* (where continuous, non-slipping rolling contact drives the robot forward). Conversely, systems that interact *with local tangential slip* rely on sliding against the tissue to generate propulsion, which we group under *sliding-based thrust*. For the *non-wall-mediated* category, the primary mechanism is *growth/eversion*, where the robot advances by physically extending or unfurling its own body rather than pushing off its environment.

Finally, we classify these systems by their *thrust-aid features*. These secondary mechanisms—which include *directional anisotropy*, *normal-force modulation*, *friction-coefficient modulation*, *temporal sequencing*, and *passive compliance*—do

not define the core locomotion class. However, they are often the critical engineering features that make the robot actually succeed in the slippery, complex environment of the colon. By structuring the review this way, we organize the devices first by how they fundamentally move, and then by the features that optimize that movement. The complete classification framework is mapped out in Figure 5.2.

## 5.4. Wall-Mediated Locomotion

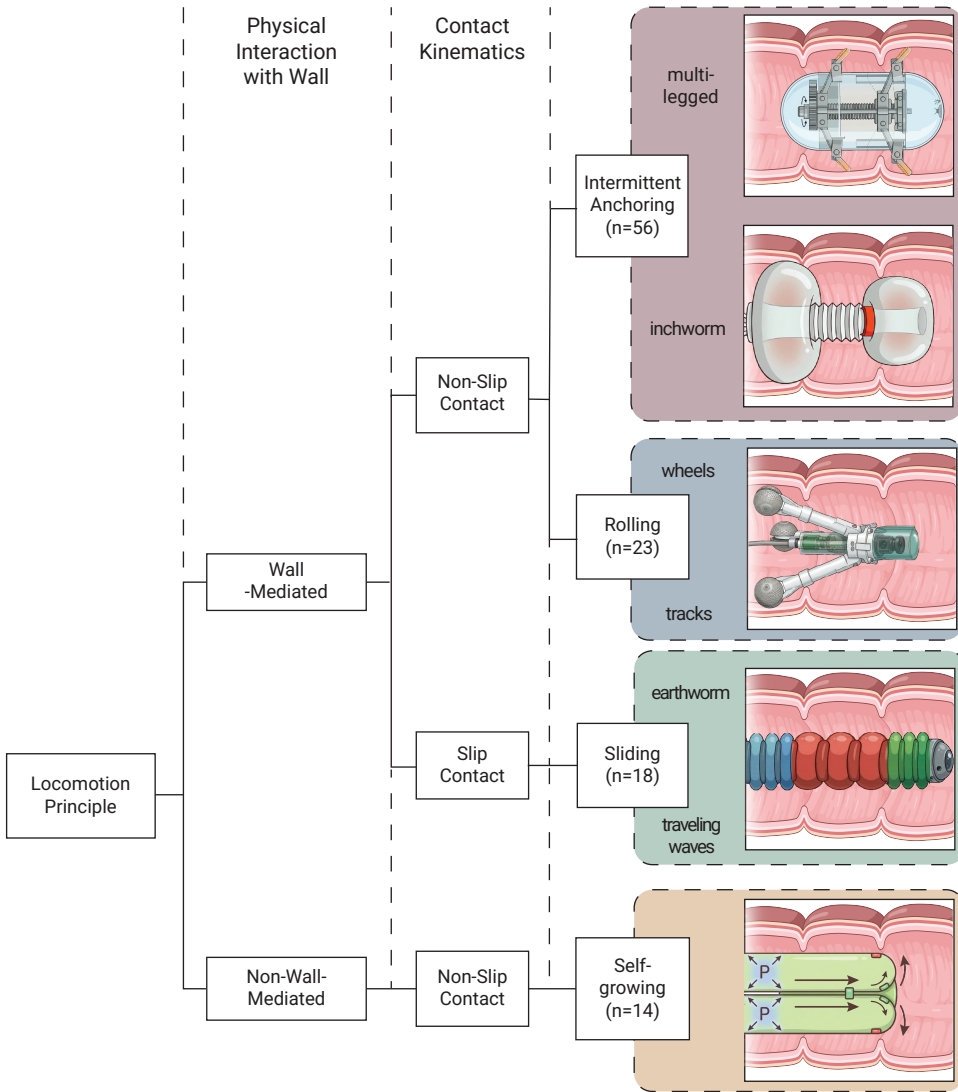
Wall-mediated locomotion includes mechanisms that rely on creating traction against the colon wall to move forward. Within this category, systems are divided based on how their moving parts interact with the tissue—specifically, whether they grip the wall firmly or intentionally slide against it.

### 5

#### 5.4.1. Locomotion Without Local Tangential Slip

Mechanisms that operate without local tangential slip—meaning their contact points do not slide or skid against the tissue while pushing the robot forward—generate propulsion by maintaining a firm, static grip. This approach is mostly seen in intermittent anchoring or continuous rolling. Intermittent anchoring systems sequentially grab and release the colon wall using discrete contact elements. Numerous designs use folding legs for this purpose. Quirini et al [24, 25] developed multi-legged devices where synchronized opening and closing of the legs produce strides without local slip, achieving forward thrust while preventing tissue damage. Similar legged designs were presented by Li et al [26], where front and rear legs alternate expanding and folding to propel the device. A widely used intermittent anchoring strategy is the inchworm-inspired mechanism, which uses one or more elongating units paired with anchoring units. These anchors secure themselves to the tissue through various methods, including expandable legs [27–51], air or fluid inflation [52–68], and vacuum suction combined with mechanical grasping, as seen in the Endotics System [69–73]. Multi-segment designs, such as those by Alcaide et al [74] and Bernth et al [75, 76], use coordinated bending, linear extension, and shape-memory alloys [77–81] to sequentially secure and advance different sections of the robot. Other intermittent approaches include using cyclical paddling strokes [82].

Rolling mechanisms, on the other hand, generate traction continuously by shifting the contact point along the colon wall without sliding. This is frequently done using continuous tracks, wheels, or helical structures. Track-based designs, such as



**Figure 5.2. Classification tree of the self-propelling locomotion principles.** The tree shows the different locomotion principles classified by their physical interaction with the wall, contact kinematics at the interface and the locomotion mechanism. with the corresponding number of collected articles per each branch.

those by Consumi et al [83, 84], Lee et al [85, 86], and others [87–91], use elastic or toothed continuous tracks to maximize contact area and friction while adapting to the width of the colon. Wheel-based systems, including those by Norton et al [92] and Bourbakis et al [93], often include expanding units to press the wheels firmly



against the wall. Accoto et al [94] demonstrated a unique rolling approach where counter-rotating wheels pinch the tissue, increasing both contact area and grip to drive the robot. Rolling contact can also be achieved through rotating flexible parts, as seen in the paddle design by Osawa et al [95] and the rotating leg motion proposed by Kim et al [96]. Furthermore, rotating the entire device body or specific spiral modules offers another non-slip continuous propulsion method, used in designs by Kim and Kim [97], Trovato et al [98], Kurata et al [99], and Liu et al [100]. Finally, continuous rolling principles have also been applied using pneumatic inverted sleeves [101] and independently driven rolling stents [102].

### 5.4.2. Locomotion With Local Tangential Slip

Mechanisms that operate with local tangential slip generate propulsion by sliding against the colon wall. Because they lack fixed anchor points, these systems rely heavily on actively changing their friction. One main method is moving a wave of friction along the segments of the device, inspired by how earthworms and snails move. By ensuring the sliding segments have less friction than the stationary segments, these multi-segmented robots move forward. Wang et al [103] and several others [47, 49, 97, 104–110] achieved this using independently controlled linear or motorized actuators to create a continuous sequence of expansion and contraction across the robot. Soft robotic versions of this traveling wave principle, powered by fluid or pneumatic actuation, have also been developed by Chan et al [111], Xin et al [112], and Saxena et al [113] to glide along the mucus-lined walls. Another approach to slip-based propulsion relies on directional friction. Yu et al [114] introduced a soft crawling robot driven by tendons that bends to selectively press parts of its body against the colon wall. This bending creates a friction difference where it is easier for the robot to slide forward than backward. Similarly, Dodou et al [115] explored the use of mucoadhesive films to selectively alter sliding friction. Finally, some slip-based systems rely on external electromagnetic fields to drag or steer the device along the tissue surface, utilizing the natural slip of the mucosa to advance smoothly, as demonstrated by Chi et al [116] and Li et al [117].

### 5.4.3. Non-Wall-Mediated Locomotion

Mechanisms that operate without relying on direct traction against the colon wall mostly include self-growing or fluid-driven systems. These devices use air or liquid pressure to propel the instrument forward, naturally following the curves of the

colon without causing significant shear stress on the mucosa. In fluid-driven and pneumatic-propulsion designs, a sealed environment can be created using the colon itself, as demonstrated by the Aer-O-Scope [118–122] which uses carbon dioxide inflation for propulsion, or hydraulic systems using an anal port [123]. Alternatively, a seal can be formed using an expandable balloon, where the inflation pressure physically pushes the tip of the colonoscope forward, a strategy used by Dehghani et al [124] and several other researchers [125–127]. Fluid-driven propulsion can also be achieved without sealing the colon entirely; for instance, Coleman et al [128] developed a highly flexible capsule propelled and steered entirely by an array of hydro-jets in a water-filled lumen.

Self-growing robots (often referred to as “vine” robots) represent another major non-wall-mediated approach. These robots advance by everting a thin, flexible material from their tip under internal pressure, meaning the robot’s body remains entirely static relative to the colon wall during insertion. Recent advancements have heavily focused on integrating active steering and clinical functionality into these soft platforms. Shi et al [129] and Kim et al [130] developed soft everting endoscopes that utilize pneumatic artificial muscles to provide omni-directional tip steering and camera stabilization. Furthermore, to address the challenge of safely retracting these flexible structures without buckling or bunching, Giri et al [131] introduced an internal, inchworm-inspired pneumatic device that acts as a steerable backbone to guide both the bending and the controlled withdrawal of the growing robot.

## 5.5. Thrust Enhancement Mechanisms

The main propulsion methods described above are often supported by secondary thrust-aid mechanisms, which help stabilizing and improving movement in the complex gastrointestinal tract. Directional friction is often built into the robots to naturally guide movement. For instance, the coated teeth on continuous tracks [84], micro-patterned treads [89], or the positive angle of spiral ridges [97] physically catch the tissue to prevent backward sliding while moving smoothly forward. Controlling how hard the robot presses against the wall (normal-force modulation) is critical across both slip and non-slip mechanisms to secure enough grip. Systems actively manage this by deploying expanding legs [27], inflating anchoring balloons and specialized chambers [52, 55, 83], expanding track tips [88], or mechanically pinching the tissue [94] to increase outward pressure before moving. Friction control is naturally

required in traveling wave designs [103], where the system relies on the colon’s natural mucus to allow contracting segments to slide while expanded segments hold firm. Furthermore, successful movement is closely tied to precise timing. The coordinated sequence of moving parts—whether it is the synchronized strides of multi-legged robots [24], the programmed activation of shape-memory springs [74], or the continuous traveling wave of earthworm-inspired segments [53]—ensures steady forward progress. Finally, passive flexibility and active steering serve as a fundamental stabilizing feature. Using elastic tracks for self-centering [85, 91], flexible rotating paddles [95], and soft silicone or fabric bodies [76, 111] allows these devices to bend with the twisted shape of the colon, maintaining the necessary contact for grip while minimizing the risk of tissue injury. To further assist with stability and safety, newer soft robots also use internal guides to smoothly retract the device [131] or use active bending to navigate tight corners without pushing too hard on the delicate tissue [117, 129].

### 5.6. Performance Evaluation and Clinical Viability

Assessing the viability of self-propelling colonoscopic robots requires a structured progression from controlled laboratory environments to complex biological systems. As devices advance through these evaluation stages, the performance metrics naturally shift from maximizing raw locomotion velocity to ensuring navigational reliability, overcoming physiological forces, and minimizing patient trauma. An overview of the specific velocities, dimensions, and evaluation stages for the reviewed systems is summarized in Table 5.2.

**Table 5.2.** Summary of reported size and performance metrics of self-propelling robotic capsules for colonoscopy categorized by locomotion mechanism.

| Reference                                                                               | Year | Size (mm) | Actuation | Stage | Vel: Avg (Max) (mm/s) |
|-----------------------------------------------------------------------------------------|------|-----------|-----------|-------|-----------------------|
| <b>Wall-Mediated: Locomotion Without Local Tangential Slip (Intermittent Anchoring)</b> |      |           |           |       |                       |
| Ren et al [52]                                                                          | 2025 | 25ø       | Pneumatic | 1     | - (-)                 |
| Ren et al [53]                                                                          | 2025 | 25ø       | Pneumatic | 1     | - (-)                 |
| Wang et al [27]                                                                         | 2023 | 30x14ø    | DC motor  | -     | - (-)                 |
| Tirado et al [54]                                                                       | 2023 | 40x11ø    | Pneumatic | 1     | 0.3 (-)               |
| Sun et al [28]                                                                          | 2022 | 27.5x16ø  | DC motor  | -     | - (-)                 |
| Chen et al [55]                                                                         | 2022 | 115x20ø   | Pneumatic | 2     | - (-)                 |

Continued on next page

Table 5.2 – continued from previous page

| Reference            | Year | Size (mm)               | Actuation     | Stage     | Vel: Avg (Max) (mm/s) |
|----------------------|------|-------------------------|---------------|-----------|-----------------------|
| Zhang et al [29]     | 2021 | 39x13 $\varnothing$     | DC motor      | 1         | 1.3 (-)               |
| Kang et al [30]      | 2021 | -                       | DC motor      | 2         | 16.9 (24)             |
| Alshorman et al [77] | 2021 | 21x15 $\varnothing$     | SMA           | 1         | - (-)                 |
| Wang et al [31]      | 2020 | 22.8x15 $\varnothing$   | DC motor      | 3         | 0.92 (-)              |
| Wang et al [32]      | 2019 | 6.5x15 $\varnothing$    | DC motor      | 1 / 2     | 1.21 / 0.94 (-)       |
| Lee et al [33]       | 2019 | -                       | DC motor      | 2         | 9.55 (-)              |
| Gao et al [34]       | 2019 | 33x24 $\varnothing$     | DC motor      | 1         | 2.63 (-)              |
| Gao et al [35]       | 2019 | 46.8x14 $\varnothing$   | DC motor      | 1 / 2     | - / 1.05 (-)          |
| Chen et al [56]      | 2019 | 95x22 $\varnothing$     | Pneumatic     | 1 / 2     | 12 / 1 (-)            |
| Manfredi et al [57]  | 2019 | 60x18 $\varnothing$     | Pneumatic     | 1         | 2.8 (-)               |
| Zhang et al [58]     | 2018 | 20 $\varnothing$        | Pneumatic     | 1         | - (-)                 |
| Yue et al [36]       | 2018 | 35x15 $\varnothing$     | DC motor      | -         | - (-)                 |
| Lee et al [37]       | 2018 | 54x16 $\varnothing$     | Stepper motor | 1 / 2     | 49.9 / 21.1 (-)       |
| Gao et al [38]       | 2017 | 38x12 $\varnothing$     | DC motor      | 2         | - (-)                 |
| Gao et al [39]       | 2017 | 20.4x14 $\varnothing$   | DC motor      | 2         | 1.1 (-)               |
| Bernth et al [75]    | 2017 | 50x35 $\varnothing$     | DC motor      | 1         | 1.21 (-)              |
| Alcaide et al [74]   | 2017 | 139x24 $\varnothing$    | SMA           | 1         | 0.42 (-)              |
| Poon et al [59]      | 2016 | -                       | Pneu/Hyd      | 1 / 2     | - (-)                 |
| Heung et al [60]     | 2016 | 100x15 $\varnothing$    | Pneumatic     | 1 / 2     | 1.5 / - (-)           |
| Gao et al [40]       | 2016 | 42x13 $\varnothing$     | DC motor      | 2         | 0.93 (-)              |
| Shi et al [41]       | 2015 | 31x16 $\varnothing$     | DC motor      | 2         | 0.38 (-)              |
| Joe et al [42]       | 2015 | -                       | DC motor      | 1 / 2     | 2.13 / 2.39 (-)       |
| He et al [43]        | 2015 | 45x14 $\varnothing$     | DC motor      | 1 / 2     | 2.9 / 0.7 (-)         |
| Chen et al [44]      | 2014 | 45x16 $\varnothing$     | DC motor      | 2         | 0.38 (-)              |
| Yanagida et al [62]  | 2013 | -                       | Pneumatic     | 1         | 3.9 (-)               |
| Lin et al [45]       | 2013 | 90x13 $\varnothing$     | DC motor      | 1 / 2     | 1.1 / 0.5 (-)         |
| Ke et al [46]        | 2013 | 82.9x15.2 $\varnothing$ | DC motor      | 1 / 2     | 1.63 / 0.13 (-)       |
| Zarrouk et al [47]   | 2012 | -                       | DC motor      | -         | 8 (-)                 |
| Park et al [61]      | 2012 | 110x9 $\varnothing$     | Pneumatic     | 1 / 2 / 3 | 56 / 33 / 8.5 (-)     |
| Nakamura et al [63]  | 2012 | 210x23 $\varnothing$    | Pneumatic     | 1 / 2     | 37.6 / 2.46 (-)       |
| Wang et al [48]      | 2012 | 98x10 $\varnothing$     | Stepper motor | 1 / 2     | 1.43 (-)              |
| Zarrouk et al [49]   | 2011 | -                       | DC motor      | -         | - (-)                 |
| Gao et al [64]       | 2011 | 105x20 $\varnothing$    | Pneumatic     | 1 / 2     | - / 0.03 (0.58 / -)   |
| Adachi et al [65]    | 2011 | 150x15 $\varnothing$    | Pneumatic     | 1         | 17.1 (-)              |
| Nakazato et al [66]  | 2010 | -                       | Hydraulic     | 1         | 1.35 (-)              |
| Kim et al [82]       | 2010 | -                       | Wired         | 1 / 2 / 3 | 10 / 6.25 / 2.83 (-)  |
| Cosentino et al [69] | 2009 | -                       | Pneumatic     | 2 / 4     | - (-)                 |
| Hosokawa et al [70]  | 2009 | 48x16x11                | SMA / Suction | 1         | 0.13 (0.16)           |
| Park et al [50]      | 2006 | 13 $\varnothing$        | DC motor      | 2 / 3     | 6.42 / - (-)          |
| Kim et al [72]       | 2006 | 148x24 $\varnothing$    | Pneumatic     | 2 / 3     | 3.16 / - (-)          |

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Table 5.2 – continued from previous page

| Reference                                                                           | Year | Size (mm)             | Actuation          | Stage | Vel: Avg (Max) (mm/s) |
|-------------------------------------------------------------------------------------|------|-----------------------|--------------------|-------|-----------------------|
| Karagozler et al [78]                                                               | 2006 | -                     | SMA                | 1     | 0.5 (-)               |
| Zuo et al [51]                                                                      | 2005 | 120x7.5 $\varnothing$ | DC motor           | 1     | - (-)                 |
| Menciassi et al [81]                                                                | 2004 | -                     | -                  | 1     | - (-)                 |
| Kim et al [71]                                                                      | 2003 | 137x24 $\varnothing$  | Pneumatic          | 2 / 3 | 2.21 / - (-)          |
| Menciassi et al [73]                                                                | 2002 | 65.5x18 $\varnothing$ | Pneumatic          | -     | - (-)                 |
| Ng et al [67]                                                                       | 2000 | - $\varnothing$       | Pneumatic          | 1 / 2 | 2.5 (-)               |
| Slatkin et al [68]                                                                  | 1995 | -                     | Pneumatic          | 1 / 3 | - (-)                 |
| Reynaerts et al [79]                                                                | 1995 | 95x15 $\varnothing$   | SMA                | -     | - (-)                 |
| Ikuta [80]                                                                          | 1988 | 215x13 $\varnothing$  | SMA                | -     | - (-)                 |
| <b>Wall-Mediated: Locomotion Without Local Tangential Slip (Continuous Rolling)</b> |      |                       |                    |       |                       |
| Wang et al [87]                                                                     | 2025 | 38x32                 | DC motor           | 1     | 30.2 (-)              |
| Du et al [88]                                                                       | 2025 | - $\varnothing$       | Ext. Drive / Track | 1     | - (29.3)              |
| Consumi et al [83]                                                                  | 2024 | 60x35 $\varnothing$   | DC motor / Pneu    | 1     | 2.8 (-)               |
| Consumi et al [84]                                                                  | 2023 | 65 $\varnothing$      | DC motor           | 1     | 4.7 (-)               |
| Liu et al [100]                                                                     | 2021 | 195x22 $\varnothing$  | Pneumatic          | 1     | 20 (-)                |
| Formosa et al [89]                                                                  | 2020 | 44x14 $\varnothing$   | DC motor           | 2 / 3 | - (40.0)              |
| Osawa et al [95]                                                                    | 2020 | 40x26 $\varnothing$   | DC motor           | 1     | - (23)                |
| Nagase et al [90]                                                                   | 2018 | 44x16.4 $\varnothing$ | DC motor           | 1     | 5.5 (-)               |
| Norton et al [92]                                                                   | 2016 | 95x38 $\varnothing$   | DC motor           | 1     | 4.9 (-)               |
| Lee et al [85]                                                                      | 2016 | 130x55 $\varnothing$  | DC motor           | 2 / 3 | 5 / - (-)             |
| Kim et al [86]                                                                      | 2014 | 32x15 $\varnothing$   | DC motor           | 1 / 2 | 24.4 / 23.8 (-)       |
| Kim et al [91]                                                                      | 2013 | 35x40 $\varnothing$   | DC motor           | 1 / 2 | 17.0 (-)              |
| Accoto et al [94]                                                                   | 2012 | -                     | DC motor           | 1     | - (145)               |
| Trovato et al [98]                                                                  | 2010 | 170x30 $\varnothing$  | DC motor           | 2 / 3 | 2.24 / 0.18 (-)       |
| Kurata et al [99]                                                                   | 2010 | 220x10 $\varnothing$  | DC motor           | 1     | 2.6 (-)               |
| Kim et al [97]                                                                      | 2010 | 78x15.6 $\varnothing$ | DC motor           | 2     | 2.7 (-)               |
| Bourbakis et al [93]                                                                | 2010 | 27x13 $\varnothing$   | DC motor           | -     | - (-)                 |
| Quirini et al [25]                                                                  | 2008 | 40x12 $\varnothing$   | DC motor           | 2 / 3 | 0.68 / 0.5 (-)        |
| Shike et al [101]                                                                   | 2008 | - $\varnothing$       | Pneumatic          | 3     | - (-)                 |
| Quirini et al [24]                                                                  | 2007 | 25x11 $\varnothing$   | DC motor           | -     | - (-)                 |
| Li et al [26]                                                                       | 2006 | 35x15 $\varnothing$   | SMA                | 2     | 0.4 (-)               |
| Breedveld et al [102]                                                               | 2004 | -                     | -                  | 1     | - (-)                 |
| Kim et al [96]                                                                      | 2002 | 125x30 $\varnothing$  | DC motor           | 1 / 2 | 1.6 / 1 (-)           |
| <b>Wall-Mediated: Locomotion With Local Tangential Slip</b>                         |      |                       |                    |       |                       |
| Yu et al [114]                                                                      | 2022 | 148x28x20             | DC motor           | 1     | - (-)                 |
| Khan et al [104]                                                                    | 2022 | 30x8 $\varnothing$    | DC motor           | 1     | 15.6 (-)              |

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Table 5.2 – continued from previous page

| Reference                                         | Year | Size (mm)              | Actuation       | Stage | Vel: Avg (Max) (mm/s) |
|---------------------------------------------------|------|------------------------|-----------------|-------|-----------------------|
| Li et al [117]                                    | 2021 | 28x14 $\emptyset$      | Electromagnetic | 1 / 2 | 7.0 (16.0)            |
| Xin et al [112]                                   | 2020 | 70x28x15               | Pneumatic       | 1 / 2 | - (0.42 / 0.03)       |
| Saxena et al [113]                                | 2020 | 50.8 $\emptyset$       | Fluid / Pneu    | 1     | - (-)                 |
| Chan et al [111]                                  | 2018 | 30x40x27               | Pneumatic       | 1     | 0.1 (-)               |
| Wang et al [103]                                  | 2017 | 105x13 $\emptyset$     | DC motor        | 1 / 3 | 2.2 / 1.9 (- / 2.2)   |
| Zarrouk et al [47]                                | 2012 | -                      | DC motor        | -     | 10.5 (-)              |
| Zarrouk et al [49]                                | 2011 | -                      | DC motor        | -     | 10 (-)                |
| Kim et al [97]                                    | 2010 | -                      | DC motor        | 2     | 0.45 (-)              |
| Ye et al [105]                                    | 2008 | 108x13.5 $\emptyset$   | DC motor        | 1 / 2 | 1.31 / 1.01 (-)       |
| Ye et al [106]                                    | 2008 | 120.5x12.1 $\emptyset$ | DC motor        | 2     | 0.93 (-)              |
| Wang et al [107]                                  | 2008 | 190x10 $\emptyset$     | DC motor        | 2     | 0.22 (-)              |
| Yan et al [108]                                   | 2007 | 120.5x12.1 $\emptyset$ | DC motor        | 1 / 2 | 5 / 0.83 (-)          |
| Wang et al [109]                                  | 2007 | 110x10 $\emptyset$     | Linear motor    | 2     | 2.56 (-)              |
| Dodou et al [115]                                 | 2005 | -                      | -               | 1     | - (-)                 |
| Yan et al [110]                                   | 2003 | 72x8.8 $\emptyset$     | Stepper motor   | 1     | 4 (-)                 |
| Chi et al [116]                                   | 2003 | 64x7 $\emptyset$       | Electromag.     | 1     | 12.88 (-)             |
| <b>Non-Wall-Mediated: Self-Growing Locomotion</b> |      |                        |                 |       |                       |
| Kim et al [130]                                   | 2026 | - $\emptyset$          | Pneumatic       | 1 / 3 | - (-)                 |
| Shi et al [129]                                   | 2025 | 18 $\emptyset$         | Pneumatic       | 1     | - (-)                 |
| Giri et al [131]                                  | 2025 | 116x26 $\emptyset$     | Pneumatic       | 1     | 1.03 (-)              |
| Manoj et al [125]                                 | 2022 | -                      | Pneumatic       | -     | - (-)                 |
| Coleman et al [128]                               | 2021 | 32x17 $\emptyset$      | Hydro-jet       | 1     | 5.3 (-)               |
| Lee et al [126]                                   | 2020 | -                      | Pneumatic       | -     | - (-)                 |
| Dehghani et al [124]                              | 2017 | -                      | Pneumatic       | 2     | 28 (33)               |
| Gluck et al [122]                                 | 2016 | -                      | Pneumatic       | 4     | - (-)                 |
| Coleman et al [123]                               | 2016 | -                      | Hydraulic       | 2     | 9.48 (-)              |
| Pourghodrat et al [127]                           | 2014 | -                      | Pneumatic       | 1     | 86.6 (-)              |
| Rex [118]                                         | 2008 | -                      | Pneumatic       | 2 / 4 | 3.18 / - (-)          |
| Arber et al [119]                                 | 2007 | -                      | Pneumatic       | 2 / 4 | - (-)                 |
| Vucelic et al [120]                               | 2006 | -                      | Pneumatic       | 4     | - (-)                 |
| Pfeffer et al [121]                               | 2006 | -                      | Pneumatic       | 3     | - (-)                 |

### 5.6.1. Benchtop Models and Phantoms (Evaluation Stage 1)

The initial evaluation of most robotic colonoscopes occurs in rigid pipes, flexible tubes, or synthetic phantoms. Devices typically record their highest locomotion velocities during this stage due to the predictable friction, uniform geometry, and lack of biological interference. For instance, wheel-based systems traversing slack cloth or

rigid tracks have demonstrated extreme velocities, such as Accoto et al [94] achieving 145 mm/s. More recently, modern track-based systems [87, 88] have comfortably reached speeds of around 30 mm/s in straight pipes, while fluid-driven systems like Dehghani et al [124] reached 28 mm/s. Soft everting robots can even exceed 300 mm/s in unconstrained, open-air benchtop tests [129]. In synthetic tubes, rotating paddle designs [95] and wheel-based expanding units [92] have recorded peak velocities between 22 and 23 mm/s.

However, standard rigid pipes fail to capture the viscoelastic nature of the colon. To bridge this gap, researchers utilize lubricated latex tubes [86], curved and inclined silicone pathways [92], and anatomically accurate large intestine rubber or hydrogel phantoms [52, 57, 83, 90, 128]. Wang et al [32] noted that transitioning from a rigid pipe to a flexible, slippery intestine inherently reduces locomotion velocity due to a loss of effective traction, underscoring the limitations of Stage 1 testing.

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### 5.6.2. Ex Vivo and In Vitro Animal Models (Evaluation Stage 2)

To test mechanisms against realistic tissue compliance, friction, and moisture, evaluations progress to excised animal colons, typically porcine. Tissues are utilized rapidly after excision to preserve their biomechanical properties. In these environments, average velocities usually decrease significantly as the robots must navigate collapsing luminal walls without causing trauma. Continuous track mechanisms [85, 86] and robust inchworm devices [67, 71, 72] generally maintain consistent speeds of approximately 2.5 to 5.0 mm/s, though some highly optimized elastic caterpillar systems have achieved speeds up to 17 mm/s ex vivo [91]. Helical rolling devices [97, 99] and various sliding-contact or electromagnetic mechanisms [74, 75, 117] typically operate in the 0.4 to 7.0 mm/s range.

Beyond locomotion speed, this stage is critical for assessing fundamental safety. Colons are routinely examined post-experiment to confirm the absence of tissue damage or mechanical marks [33]. Furthermore, safety protocols such as thermal testing are conducted to ensure internal actuators or power sources do not induce heat injury to the intestinal wall [34]. Quirini et al [25] enhanced this stage by mounting ex vivo tissue onto a fixture mimicking the human mesentery, allowing for accurate anatomical testing of the device's ability to negotiate natural flexures.

### 5.6.3. In Vivo Animal Trials (Evaluation Stage 3)

In vivo trials introduce the active physiological challenges of a living organism, most notably peristalsis (the natural muscular contractions of the gut), abundant fluid, and fecal matter. These factors dramatically increase the difficulty of forward propulsion. For example, during in vivo tests in a living pig, the multi-legged device by Quirini et al [25] advanced at 0.5 mm/s while actively resisting peristaltic forces, and Trovato et al [98] recorded an average velocity of 0.18 mm/s with their dual-helical device. Traveling-wave devices and shape-morphing soft robots have shown success here, with Wang et al [103] and Bernth et al [76] achieving speeds around 1.6 to 2.3 mm/s depending on the instantaneous curvature of the living colon. Conversely, powerful micro-treaded capsules [89] and soft growing robots [130] have demonstrated the ability to more rapidly bypass these obstacles in animal models.

As noted by Lee et al [85], navigating biological debris remains a primary hurdle. Consequently, post-operative endoscopic examinations are standard practice at this stage to rigorously verify that the active propulsion mechanisms did not compromise mucosal integrity, a safety benchmark successfully met by systems like the ColonoSight before advancing to human use [101].

### 5.6.4. Human Clinical Trials (Evaluation Stage 4)

When self-propelling robots reach human clinical trials, the primary performance metrics transition entirely away from raw velocity and focus heavily on clinical outcomes and patient experience. The ultimate goal of biomimetic locomotion is to eliminate the tissue stretching and looping associated with conventional, pushed endoscopes. Cosentino et al [69] evaluated the inchworm-inspired Endotics System in human patients, explicitly comparing it to standard colonoscopy. Similarly, Shike et al [101] validated the disposable, power-assisted ColonoSight system across hundreds of patient procedures, proving it could safely navigate the human colon without relying on a stiff pushing shaft. In this final evaluation stage, success is quantified by diagnostic accuracy, the time required to reach the cecum, overarching safety, and significantly reduced patient pain and discomfort.



## 5.7. Discussion

The comprehensive review of self-propelling mechanisms for colonoscopy reveals clear trends in morphological design, actuation strategies, and performance characteristics. By analyzing these systems across their fundamental locomotion principles and evaluation stages, several critical insights emerge regarding the current state of the art and the physiological challenges that remain.

### 5.7.1. Morphological Constraints and Size Trends

An analysis of the dimensional data highlights a strong correlation between a robot's primary locomotion principle and its geometric profile. Mechanisms relying on continuous rolling without local tangential slip [85, 92] frequently exhibit larger radial diameters, often exceeding 35 mm and reaching up to 65 mm. This larger volume is necessary to house the tracks, wheels, or expanding paddle units required to maintain continuous, non-slip contact with the luminal wall. Conversely, systems employing sliding-based thrust—particularly those utilizing traveling waves [103, 105, 107]—are distinctly slender. These devices frequently measure less than 15 mm in diameter but exceed 100 mm in length. This high aspect ratio accommodates the multiple segments required to generate a net positive friction difference.

Crucially, as devices progress toward clinical viability in Evaluation Stages 3 and 4, their dimensions naturally converge toward the anatomical limits of the human colon. Regardless of the underlying mechanism, systems evaluated *in vivo* or in human trials typically maintain radial diameters below 25 mm [25, 69, 71]. This reflects the absolute necessity of minimizing tissue stretching to ensure patient safety and comfort.

### 5.7.2. Performance Shifts Across Evaluation Stages

A consistent trend across the literature is the noticeable decrease in locomotion velocity as devices transition from benchtop models (Stage 1) to living biological environments (Stages 3 and 4). In rigid or idealized Stage 1 environments, rolling mechanisms generally demonstrate the highest raw speeds, with some theoretical benchtop designs recording maximums up to 145 mm/s [94]. However, because continuous rolling relies on a predictable, stable contact surface, these mechanisms experience significant efficiency reductions when placed in the collapsing, highly lubricated, and flexible environment of a real colon.

In contrast, intermittent anchoring and sliding-based mechanisms exhibit lower absolute speeds in Stage 1 testing (typically between 1.0 and 15 mm/s) but demonstrate greater resilience in ex vivo (Stage 2) and in vivo (Stage 3) environments. Because these mechanisms are designed to actively grip the tissue or intentionally utilize the mucosal lubrication to slide, their speed reduction is less pronounced in biological settings. For example, multi-legged or inchworm systems generally stabilize between 0.5 and 2.5 mm/s in living tissue [25, 71]. Self-growing devices also show strong velocity retention; because they navigate via internal pressure rather than external wall traction, they report consistent velocities across both Stage 1 [127] and advanced clinical stages [118].

### 5.7.3. Evolution of Actuation Mechanisms

A chronological assessment reveals a clear technological shift in actuation strategies over the decades. Early self-propelling colonoscopes developed in the late 1980s and 1990s heavily utilized Shape Memory Alloy (SMA) springs [79, 80]. While SMAs allowed for excellent miniaturization, their slow thermal cooling cycles limited locomotion velocities to less than 1 mm/s.

Through the 2000s and 2010s, researchers largely transitioned to DC motors and electromagnetic actuators. This allowed for more complex, coordinated movements—such as synchronous stepping or helical rotation—yielding notable improvements in speed and control [31, 99]. Recently, however, there has been a pronounced shift toward pneumatic and hydraulic actuation [55, 111, 112]. Reflecting broader trends in soft robotics, contemporary designs increasingly use fluidic networks to create compliant bodies that can gently deform to the colon's geometry, prioritizing mucosal safety over rigid mechanical power.

### 5.7.4. Identified Opportunities for Future Work

The current literature highlights several areas for future development and optimization. A notable limitation is the tendency to optimize mechanisms for rigid, controlled environments. Designing a device with continuous tracks or high radial expansion to achieve high benchtop speeds often translates to reduced effectiveness against the soft, flexible walls of a living colon.

Another major challenge is the high variability in testing methods across all the presented studies. Because researchers use different synthetic phantoms, animal

tissues, and experimental setups, it is difficult to compare performance or draw firm conclusions on which mechanism is better. A big opportunity exists to develop a unified testing and validation platform. Standardizing these environments would accelerate the research validation process and allow for more representative, direct comparisons between competing designs.

Similarly, there is a major opportunity in developing universal models to simulate the interaction between self-propelling robots and the colon tissue. Accurately predicting the traction and friction forces of a colonic robot requires a deep understanding of this tissue-device interface. Creating universal predictive tools would allow researchers to accurately simulate capsule behavior before building physical prototypes, greatly speeding up the design process.

Furthermore, the integration of a physical tether presents a specific challenge. For modern robotic capsules, a tether is highly advantageous—and often essential—for delivering continuous electrical power, transmitting high-definition video, and providing a working channel for instrument transfer (such as biopsy forceps or therapeutic tools). While the cumulative friction of pulling this tether through the gastrointestinal tract can introduce parasitic drag, this effect can be effectively mitigated. The application of highly lubricious hydrogel coatings, the integration of active tether-feeding systems, and the use of ultra-compliant materials allow tethered capsules to maintain deep insertion capabilities without compromising their functional payload.

In parallel, self-everting (or growth-based) robots represent a highly promising alternative approach. Because these systems navigate by everting internal material via fluidic pressure rather than relying on external wall traction, they navigate complex anatomies with virtually zero shear stress on the tissue. While current iterations face structural challenges in housing large working channels for clinical tools, the fundamental principle of growth-based deployment offers an atraumatic approach that could significantly inform future hybrid designs.

Ultimately, for tethered capsule systems, sliding-contact (slip) locomotion presents the most compelling pathway to overcome the colon's physiological barriers. Rather than attempting to gain traction by pressing outward against the sensitive luminal wall—which increases patient discomfort and risks tissue trauma—slip-based mechanisms uniquely capitalize on the colon's naturally slippery, mucus-lined environment. By utilizing engineered anisotropic friction or continuous traveling waves, these devices can glide forward efficiently, utilizing the low-friction environment to help offset tether drag. The approach fundamentally reduces the normal force

required for propulsion, seamlessly adapting to the collapsing anatomy of the bowel. Consequently, further advancing slip-sliding locomotion offers a robust and highly scalable foundation for creating faster, safer, and more capable diagnostic and therapeutic colonoscopic tools.

## 5.8. Conclusion

This chapter presented a classification framework for self-propelling mechanisms for robotic colonoscopy, categorizing the state of the art into wall-mediated and non-wall-mediated mechanisms. A review of the literature indicates a notable performance gap between idealized benchtop testing and in vivo applications, as mechanisms optimized for rigid environments often face challenges navigating the dynamic, viscoelastic anatomy of the human colon.

Consequently, there is growing interest in compliant, soft-robotic systems that prioritize anatomical conformance and mucosal safety. Non-wall-mediated self-everting robots, for instance, offer an atraumatic approach by navigating via internal pressure rather than external wall traction. For tethered capsule systems, the tether remains essential for clinical utility—providing power, data, and instrument access—but it introduces drag that can limit insertion depth.

To address these physiological challenges, sliding-contact (slip) locomotion presents a practical alternative. Rather than relying entirely on static traction against delicate tissue, slip-based mechanisms utilize the colon's naturally lubricated environment to facilitate propulsion. Exploring this interaction offers a pathway to potentially reduce tissue strain and better adapt to the bowel's collapsing anatomy. This biomechanical approach establishes the underlying rationale for the slip-sliding architectures investigated in the subsequent chapters.

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# CHAPTER 6

## Self-Propelling Locomotion via Motion-Coordinated Friction Anisotropy

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## Abstract

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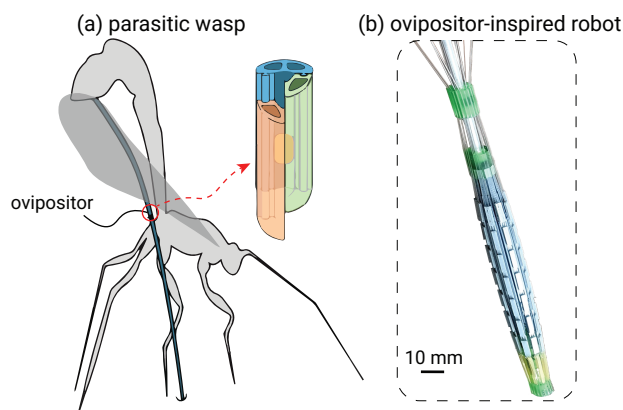
Self-propelling robots for confined spaces tune their frictional interaction with the surrounding environment to generate the directional thrust required for locomotion. However, achieving stable locomotion across environments with varying sizes, shapes, elasticity, lubricity, and loading conditions remains challenging. Building on the self-propelling locomotion principles explored in Chapter 5, this chapter investigates motion coordination as a method to actively tune the system-level friction anisotropy to achieve self-propelling locomotion ('propel'). In this chapter, we present and evaluate a bio-inspired self-propelling robot concept that generates stable thrust through the coordinated motion of multi-contact sliders. Inspired by the reciprocating motion pattern wasps use to transport eggs through their ovipositor needle, the robot sequences the motion of these sliders to create the frictional anisotropy necessary for forward locomotion. To dynamically adapt to varying environmental topographies, we use flexible sliders that radially expand through axial deformation to maintain sufficient grip on the wall. Our experiments demonstrate that the prototype successfully achieves locomotion across varying tube sizes, shapes, and payload conditions for diameter ratios ranging from 0.7 to 1.5, maintaining an average locomotion efficiency of 70% with a maximum anchoring force of 13 N. These results demonstrate the viability of motion coordination as a robust strategy to actively tune the system-level friction anisotropy, establishing the mechanical framework for the self-propelling robotic colonoscopy capsule technology in Chapter 7.

## 6.1. Introduction

Pipelines are ubiquitous mean of fluid media transport, significantly impacting various aspects of our daily lives. They fulfill a multitude of roles, from the transportation of water, gases, and oils to the essential circulation of bodily fluids. These pipelines vary widely in size, ranging from sub-millimeter dimensions within biological systems to industrial counterparts measuring tens of centimeters. They also exhibit diverse curvatures, encompassing both straight and sharply curved configurations, and possess a wide array of structural variations, from rigid to soft. Their span extends over a broad range of distances, encompassing mere centimeters to extended lengths of hundreds of meters. Despite the wide range of pipeline attributes, they all share a common necessity: regular inspections to ensure their proper functioning. For this purpose, several pipeline inspection robots have been developed targeting different applications.

Various inspection robots have been developed for pipeline navigation, encompassing wheel-type [2, 3], walking-type [4], crawler-type robots [5], screw-driven type [6] and other locomotion mechanisms [7, 8]. These robots typically feature rigid structures, powered by electric motors and gear transmission systems, making them well-suited for inspecting pipelines with larger diameters. However, these designs exhibit certain limitations, such as a lack of flexibility that hinders their adaptability to pipelines of varying sizes. Downsizing these robots, particularly for millimeter-scale pipelines as found in biological systems, poses significant challenges. Furthermore, their intricate mechanisms necessitate complex assembly structures, which are not only maintenance-intensive but also prone to failure if debris enters their joints. The utilization of soft material can alleviate these drawbacks by rendering the robot completely soft.

Soft robots have emerged as a promising technology for pipeline inspection. Their intrinsic flexibility enables them to conform to a variety of pipeline shapes and sizes, ensuring safe interaction with pipeline interiors, a particularly crucial feature in the context of biological pipelines. Most of the existing soft robotic designs for pipeline inspection rely on fluid for locomotion by harnessing a supply of air or hydraulic pressure to generate the motion, including inchworm-type [10–12], earthworm-type [13, 14] and growing-type [15] robots. Their relatively simple design offers a scalable alternative to rigid counterparts. Despite their advantages, fluidic-based designs share common drawbacks such as slower speeds and reduced durability. As robots traverse deeper into pipelines, fluidic response delays become more pronounced. Addi-



**Figure 6.1. Mechanically-inflatable bio-inspired locomotion system.** The working principle of the robot proposed herein is inspired by how a parasitic wasp uses its ovipositor to inject its eggs into a host medium. (a) illustration of the parasitic wasp, adapted from [9]. (b) prototype of the ovipositor-inspired robot proposed herein.

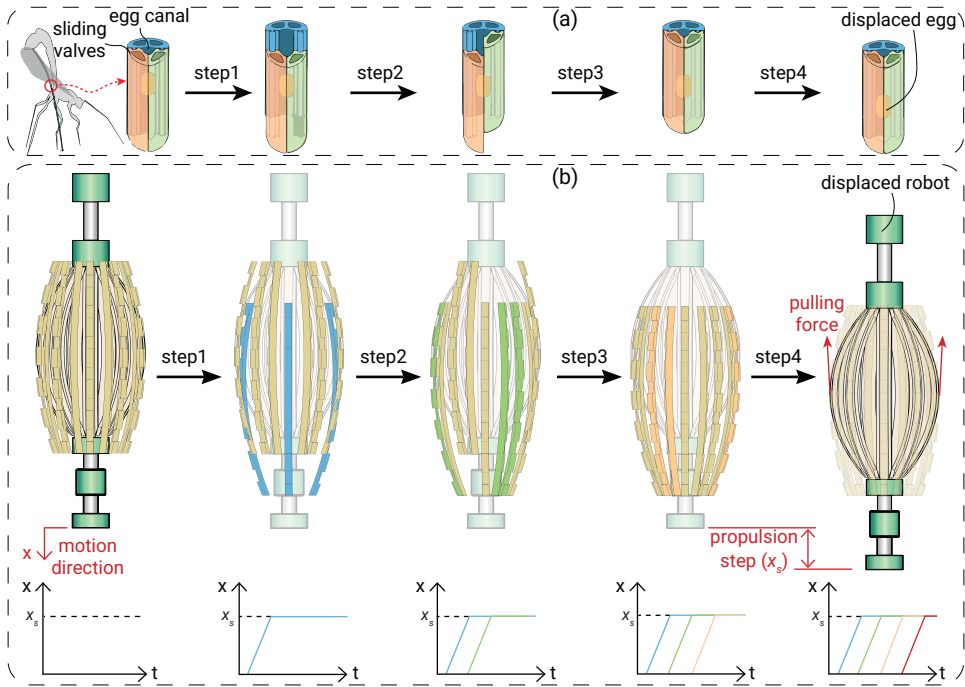
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tionally, they are susceptible to rupture, especially in industrial pipelines where debris may be present. Hence, the quest for novel solutions that retain the merits of soft robots while addressing their drawbacks becomes evident.

Inspired by nature, we propose an ovipositor-inspired flexible locomotion system, Fig.6.1, that uses reciprocating flexible sliders to propel while adapting to tubes of different shapes and sizes through a mechanical inflation technique, pertaining the advantages of the soft structure while eliminating the downfalls of fluidic-based designs. In this study, we translated this principle into a fully functional prototype which we validated and tested experimentally. The remainder of this chapter is organized as follows: in Section 6.3, we provide an overview of the working principle, detailed design and prototype development of the proposed locomotion system. In Section 6.4, we describe the experiments conducted to validate and characterize the developed prototype. Afterwards, we present and discuss the experimental results followed by a conclusion and outlook for the future in Section 6.6.

## 6.2. Bio-inspiration from Wasp Ovipositor

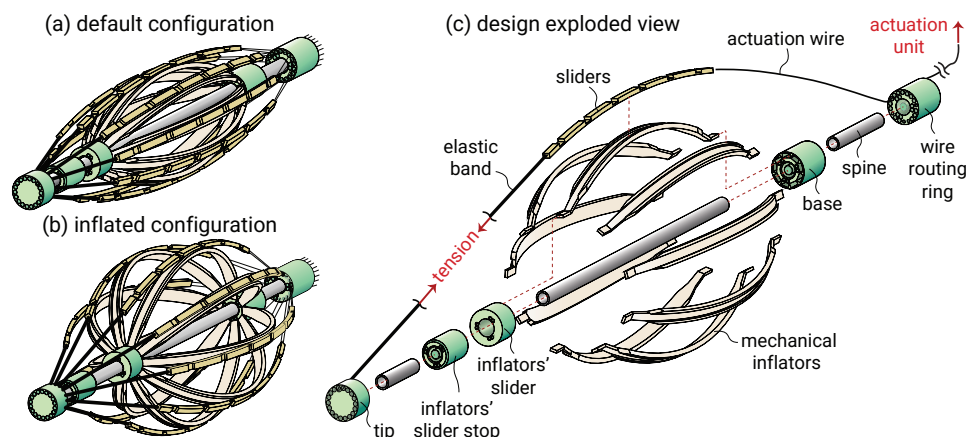
An intriguing biological example of reliable mass transport within a tubular structure is evident in parasitic wasps. These parasitic insects are equipped with an ovipositor, an exceedingly thin and flexible tubular organ enabling them to penetrate living hosts for the purpose of depositing their eggs inside this host, as illustrated in Fig.6.2(a).



**Figure 6.2. Working principle of the wasp-inspired mechanically-inflatable locomotion system.** (a) the hypothesized motion sequence of the valves that the parasitic wasp uses to transport its eggs along its ovipositor, adapted from [9]; the valves retract sequentially, with one valve retracting and the other two remaining stationary, while the egg stays in place due to the higher friction of the stationary pair (step1-3). Once all the valves have been retracted, the wasp extends the entire valve assembly simultaneously, resulting in a displacement of the egg (step4). (b) the motion sequence of the ovipositor-inspired robot; a total of nine sliders are organized into three distinct groups, each consisting of three sliders. These groups operate in a coordinated, yet independent manner. The motion sequence initiates by advancing the first group of sliders forward, while the other two groups remain stationary. This pattern is then repeated with the other two groups until all groups have advanced into the forward position (step1-3). Subsequently, all three groups are pulled simultaneously, harnessing the cumulative friction force to propel the robot body forward, assuming the cumulative friction force surpasses the body inertial force (step4) (refer to the supplementary material video.1-2 for a demonstration of the motion sequence).

The transfer of eggs through the ovipositor is accomplished via an oscillatory motion of the ovipositor valves, which collectively constitute the tubular ovipositor [9, 16–18]. The ovipositor typically comprises three valves that move independently: a dorsal valve and two ventral valves. Enveloped by an ovipositor sheath, these valves can slide longitudinally relative to each other while being secured radially through a tongue-and-groove connection [19].

The state-of-the-art leading theories suggest that the friction between the valves and the eggs enables the transportation of the eggs along the ovipositor [16, 19]. One



**Figure 6.3. Detailed design of the locomotion system.** (a) shows the locomotion system in its default (deflated) configuration. (b) shows the system in its inflated configuration. (c) shows the exploded view of the design featuring the main components of the system; the flexible sliders (a single slider is displayed for clarity), mechanical inflators and their associated actuation and structural support components (wires, sliders and rings).

## 6

theory suggests that egg transportation and drilling occurs concurrently. In this theory, the motion sequence of the valves allows the ovipositor to penetrate deeper into the substrate while simultaneously transporting and depositing the eggs. Another theory posits that wasps transport their eggs via the ovipositor independent of penetration into the substrate. In this case, repeated small relative displacements of the ovipositor valves enable the transportation of the eggs, as depicted in Fig.6.2(a).

The sequence of egg transport, according to the second theory, initiates with one of the valves retracting in the direction opposite to the intended transport direction, while the other two valves stay in place (Step 1 in Fig.6.2(a)). Consequently, the egg stays in place because the collective friction between the two stationary valves and the egg surpasses the friction exerted by the single retracting valve, provided that each valve has the same contact area with the egg. Subsequently, the second valve retracts while the other valves stay in place (step2 in Fig.6.2(a)). Then, the last valve retracts while the rest stay in place (step3 in Fig.6.2(a)). Finally, all three valves translate simultaneously in the intended transport direction (step4 in Fig.6.2(a)), where the egg advances in tandem with the valves due to friction. By this step, all valves will have returned to their initial positions, and the egg will have moved forward by one stroke length. By repeating this cycle, the egg can be transported along the entire length of the ovipositor until its deposition.

### 6.3. Design and Implementation of Wasp-Inspired Adaptable Locomotion System

Inspired by the egg transport mechanism of the wasp ovipositor, we designed a flexible self-propelling locomotion system, which comprises a wasp-inspired system of sliders to propel in tubular confined environments, integrated with a mechanical inflation system to adapt to different tube sizes and shapes. Fig.6.3(a) and (b) show the locomotion system in its default and inflated configurations respectively.

#### 6.3.1. Self-Propelling Locomotion System

In this design, the original arrangement of three sliding valves within the ovipositor case has been reconfigured into three groups of flexible sliders, as shown in Fig.6.3(a). Each group comprises three sliders, which operate collectively. This adaptation, from a single valve to a trio of sliders, enhances the robot capacity to conform to various lumen shapes. Each slider is connected to the tip through an elastic band at its distal end and is actuated through control wires attached to its proximal end, while resting on the mechanical inflators. The actuation wires are systematically organized and routed through a wire routing ring to constrain each wire to its corresponding slider. The structure of the locomotion system is held together using a rigid backbone (spine), which is a series of rigid tubes whose lumen can potentially allow for the deployment of inspection tools.

The groups of sliders follow the same motion pattern as the ovipositor valves, as illustrated in Fig.6.2(b). In the first step, one group of sliders advances forward, while the other two remain stationary. The greater static friction of the two stationary slider sets allows the moving group to progress without displacing the entire robot. This step is then repeated for the remaining two groups until all the sliders are in the forward position. Consequently, when all the slider groups are pulled simultaneously, the body moves forward by a single propulsion step ( $x_s$ ), given that the static friction of all sliders surpasses the inertial force of the system. By repeating this sequence, the robot continues to move forward. This motion pattern is achieved through a cam and sliders mechanism in the actuation unit, with a cam profile that aligns with the motion sequence. The elastic bands, connecting the flexible sliders to the tip, play a crucial role in spring-loading the cam sliders, forcing them to follow the cam profile (refer to the supplementary material video.1-2 for a demonstration of the motion sequence).

### 6.3.2. Mechanical Inflation System

The mechanical inflation mechanism comprises elastically deformable inflators affixed to a stationary end and a movable end. The movable end slides along the spine of the robot and is actuated via control wires, as depicted in Fig.6.4. When those wires are pulled, the movable end moves closer to the stationary end. Consequently, the inflators undergo a radial elastic deformation, resulting in the mechanical inflation action shown in Fig.6.4. Upon releasing of the wire tension force, the elastic energy stored within the inflators springs the movable end back, restoring its default deflated state (refer to the supplementary material video.3 for a demonstration of the mechanical inflation mechanism).

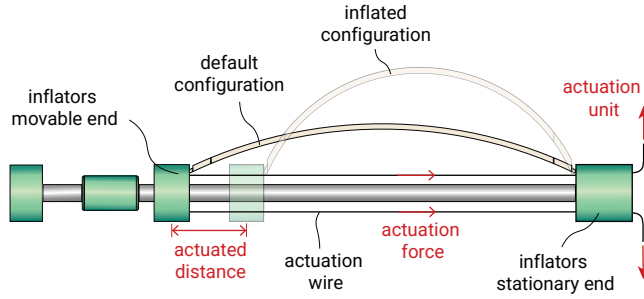
To integrate the inflation mechanism into the locomotion system, we embedded the inflators beneath the flexible sliders which need to be in contact with the environment. To ensure the alignment of the inflators and the flexible sliders, we designed the inflators to have longitudinal channels which constrain the sliders laterally while allowing them to slide freely in the axial direction, as shown in Fig.6.3(c).

### 6.3.3. Prototype Development

To validate the design, we implemented a proof-of-concept prototype. We optimized the design to allow for an extensive use of 3D printing technology for ease of fabrication. Consequently, all of the parts were 3D printed with the exception of the robot spine, for which we used off-the-shelf rigid aluminum tubes. The flexible sliders and mechanical inflators were 3D printed out of polylactic acid (PLA) filament material using a fused deposition modeling (FDM) 3D printer, Ultimaker3 (Ultimaker, Utrecht, The Netherlands). The rest of the components were printed using a resin-based stereolithography (SLA) 3D printer, Formlabs3B (Formlabs, Somerville, Massachusetts, U.S.A.). We employed off-the-shelf nylon wires to control the flexible sliders and mechanical inflators, and we used latex elastic bands to connect the robot's tip to the distal ends of the flexible sliders.

## 6.4. Experimental Validation

To validate the concept of the mechanically-inflatable self-propelling locomotion mechanism and characterize its performance, we conducted two experiments. The first experiment serves as a proof-of-concept of the capability of the locomotion



**Figure 6.4. The mechanical inflation mechanism.** The mechanical inflators are mounted to a stationary end and a movable end. By sliding the movable end along the axis of the robot, the inflator undergoes an elastic deformation, resulting in radial mechanical inflation (refer to the supplementary material video.3 for a demonstration of the mechanical inflation mechanism).

mechanism in traversing through tubes in different conditions. The second is to characterize the load-carrying capacity of the mechanical inflation mechanism.

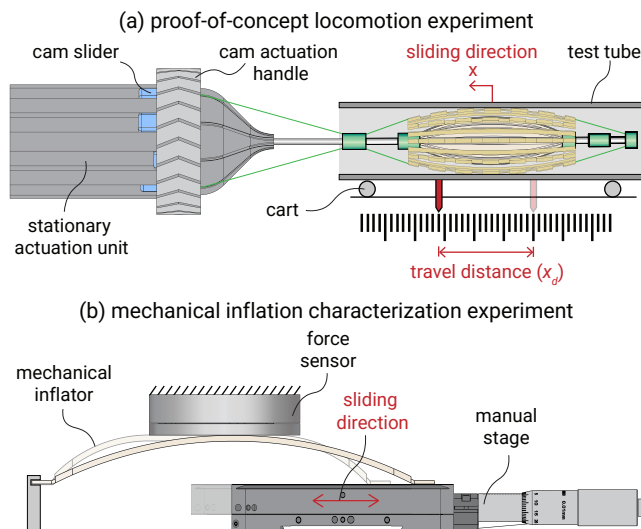
#### 6.4.1. Proof-of-Concept Experiment

In this proof-of-concept experiment, we secured the robot prototype in place and positioned the test tube on a cart, as shown in Fig.6.5(a). This arrangement allows the tube to move freely in the axial direction, while the robot itself remains stationary, minimizing operator influence on locomotion forces. To measure the travel distance of the tube, we affixed a pointer to the cart, which moved along a measuring tape, as depicted in Fig.6.5(a). To resemble the weight of the robot while sliding, we created an offset between the axis of the robot and that of the tube, which results in a side pushing force equivalent to the weight of the robot (refer to the supplementary material video.4 for a demonstration of the proof-of-concept experiment).

The experimental procedure started by positioning the robot entirely inside the tube and then recording the cart initial position. Subsequently, the robot was actuated to achieve one complete sliding cycle, after which the final position was recorded and the cart was reset to its initial position. This procedure was repeated three times for each testing condition. We conducted two sets of experiments using the same setup. First, we studied the performance of the robot across different combinations of tube sizes and robot radial inflations. Second, we studied the effect of loading, irregularity in the tube shape on the locomotion performance.

To quantify the performance of the locomotion mechanism, we introduce two variables; the locomotion efficiency ( $\eta$ ) and the diameter ratio ( $\delta$ ). We define the loco-





**Figure 6.5. Experimental setups.** (a) shows the experimental setup used to characterize the efficiency of the locomotion system in different conditions (refer to the supplementary material video.4 for a demonstration of the proof-of-concept experiment). (b) shows the experimental setup used to characterize the holding force of the mechanical inflation system (refer to the supplementary material video.5 for a demonstration of the mechanical inflation characterization experiment).

motion efficiency as the ratio between the distance traveled by the tube to the theoretical distance the tube could have traveled in case of ideal force transmission between the robot and the tube, expressed as follows:

$$\eta_l = \frac{|x_{final} - x_{initial}|}{x_{theoretical}} \times 100 \quad (6.1)$$

In the theoretical case, we assume a perfect slip between the flexible sliders and the tube when slider groups are individually advanced forward (step1-3 in Fig.6.2(b)), and no slip when slider groups are all pulled back together (step4 in Fig.6.2(b)). The diameter ratio ( $\delta$ ) is the ratio between the diameter of the robot to the diameter of the tube, expressed as follows:

$$\delta = \frac{d_{robot}}{d_{tube}} \quad (6.2)$$

|                        |     |      |      |      |      |      |
|------------------------|-----|------|------|------|------|------|
| actuated distance (mm) | 0.3 | 0.9  | 1.5  | 2.1  | 2.7  | 3.3  |
| robot diameter (mm)    | 15  | 16.7 | 18.7 | 20.5 | 21.4 | 23.1 |

**Table 6.1.** Distance-to-diameter mechanical inflation mapping.

To estimate the diameter of the robot, we measured the diameter in free condition (no contact condition) in relation to the actuated distance of the inflators' movable end depicted in Fig.6.4. This distance-to-diameter mapping is summarized in Table.6.1.

#### 6.4.2. Mechanical Inflation Characterization

In this experiment, we characterized the holding force produced by the inflation mechanism. To do so, we affixed one end of a single slider to a manual linear stage and the other end to a fixture. We then measured the normal (holding) force generated by the mechanical inflation of the slider, actuated by the manual stage, using a force sensor touching the mechanical inflator, as shown in Fig.6.5(b) (refer to the supplementary material video.5 for a demonstration of the mechanical inflation characterization experiment).

The experimental procedure started by fixing the slider to the manual stage such that the manual stage initial position is exactly where the mechanical inflation is about to occur. The force sensor is then brought in touch with the mechanical inflator such that the sensor is not loaded. We then actuate the inflator using the manual stage while recording both the normal force and the distance traveled.

### 6.5. Results and Discussion

The locomotion experiments revealed a relationship between the locomotion efficiency ( $\eta_l$ ) and the diameter ratio ( $\delta$ ) across different combinations of tube sizes and robot diameters. Our findings revealed that the locomotion efficiency did not exhibit a direct correlation with either the robot size or the tube diameter alone. Instead, we observed a positive linear correlation ( $r = 0.6434$ ) between the locomotion efficiency and their ratio, the diameter ratio, as shown in Fig.6.6(a). This positive linear correlation is intuitive since as the diameter ratio increases, more sliders come into contact with the tube, leading to an expansion of the contact area and an increase in the transmitted pulling force, which consequently results in higher locomotion efficiency. Remarkably, the robot was able to propel at diameter ratios ( $\delta < 1$ ) with an efficiency of ( $\eta_l \approx 60 - 70\%$ ), despite the robot size being smaller than the tube diameter, demonstrating its capability to propel without inflation. This result aligns with the wasp-inspired locomotion working principle, as only three of the nine sliders

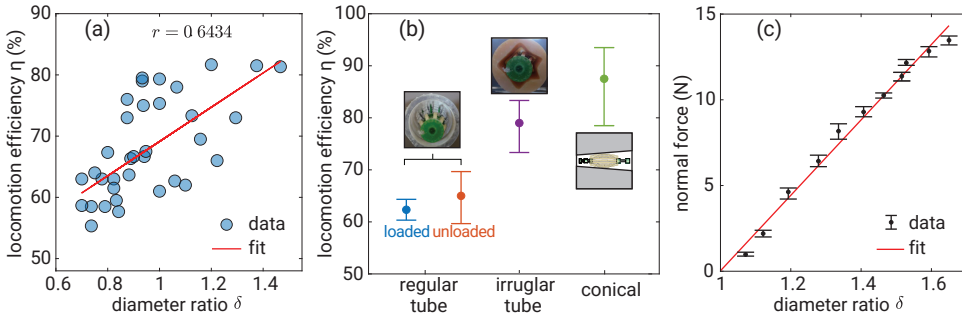
need to make contact, to propel. It is worth noting that at this range of diameter ratios ( $\delta \leq 1$ ), the robot can only locomote in a horizontal tube configuration, and will not be able to propel through a vertical configuration.

A noteworthy aspect of locomotion with a small diameter ratio ( $\delta < 1$ ) is the situation where only four or two sliders come into contact with the tube. In such instances, a specific occurrence takes place during the individual slider groups advancement steps (step1-3 in Fig.6.2(b)): at one of these steps, an equal number of sliders move forward as sliders that remain stationary. This balance can lead to the elimination of friction asymmetry, potentially hindering the locomotion efficiency. Consequently, we anticipate that there might be certain combinations of tube sizes and robot diameters where efficiency is reduced when only two or four sliders are in contact. However, this particular condition has not been prominently observed in our experiments.

The following experiment aimed to assess the effect of cart weight on the locomotion efficiency. We conducted this experiment by loading the tube with a (2 Kg) weight, and then compared the results to the unloaded condition, all at a specific diameter ratio ( $\delta = 1$ ). The added weight introduced a higher friction requirement for tube movement. Unexpectedly, the experiment revealed that the load had no significant effect on locomotion efficiency. Specifically, the robot achieved an average efficiency of (63%) under the loaded condition, as compared to (65%) in the unloaded condition, as depicted in Fig.6.6(b). This outcome can be attributed to the robot slow accelerations and the relatively low friction of the cart used in the experiment.

It is important to note that this experiment solely investigated the impact of added inertia due to the weight. However, the weight of the robot itself can influence other aspects of the locomotion mechanism, such as the distribution of contact across the sliders. In cases where the robot weight increases, the load borne by the sliders at the bottom also increases, subsequently expanding their contact area compared to the other sliders. This effect can be mitigated by activating the mechanical inflation mechanism, which elevates the normal force across all sliders, resulting in a more equal distribution of friction forces.

Furthermore, we examined the capability of the robot to navigate through arbitrarily shaped tubes, simulating scenarios where tube obstructions might occur, such as due to the accumulation of contaminants. The robot demonstrated successful propulsion, achieving an average locomotion efficiency of (79%) for the specific arbitrary shape illustrated in Fig.6.6(b). Our examination was not exhaustive and was only intended to offer a preliminary indication of the robot performance in such situations.



**Figure 6.6. Experimental results.** (a) shows the correlation between the locomotion efficiency ( $\eta$ ) of the robot and the diameter ratio ( $\delta$ ). (b) shows the locomotion efficiency ( $\eta$ ) of the robot in tubes of different shapes and at different payloads. (c) shows the normal (holding) force of a single mechanical inflator as a function of the diameter ratio.

Consequently, efficiency is expected to vary significantly depending on the specific tube shape. However, we anticipate that the inflation mechanism will mitigate potential performance limitations by conforming to the lumen shape.

Contrary to our expectations, the experiment involving a conical tube yielded favorable results. We had anticipated that the conical shape would exert a counteractive force on the robot due to its shape, hindering the intended movement and reducing efficiency. Instead, the system exhibited the most favorable locomotion efficiencies among all tests. This unexpected outcome could be attributed to the increasing diameter ratio ( $\delta = 1.1 \sim 1.46$ ) along the length of the conical tube, which increases the contact area of the sliders as the robot propel inside the tube, resulting in an overall increase in the locomotion efficiency. This suggests that the system performs optimally when it has a larger contact area, which implies its potential effectiveness in a soft pipeline environment where the surroundings can conform to the robot shape.

The mechanical inflation characterization experiment revealed a remarkable load carrying capacity for the inflation mechanism. Our findings indicated that a single inflator exhibited a stiffness of ( $\approx 4.2$  N) per (1 mm) of actuation distance. With as little as 3mm of actuation, the robot could achieve a diameter ratio ( $\delta = 1.6$ ) and a holding force of ( $\approx 13$  N). In a hypothetical scenario where the robot is propelling in a vertical tube, the maximum load it can carry is determined by the cumulative frictional forces generated by its flexible sliders. These frictional forces can be estimated by multiplying the applied normal (holding) force with the static coefficient of friction ( $\mu_s$ ) between the flexible sliders and the tubular environment. Assuming a coefficient of friction of ( $\mu_s = 0.5$ ), the robot has the capacity to transport approxi-

ately (5.8 Kg) of payload, which includes its own weight. This finding underscores the robot's ability to excel in a vertical tube configuration, a scenario demanding robust load-carrying capacity to counteract the robot's own weight and potential occlusion-induced forces. Furthermore, the ability to sustain high forces expands the range of tasks the robot can undertake, especially those involving force exertion.

## 6.6. Conclusion and Future Work

In this chapter, we introduced a novel concept of an ovipositor wasp-inspired flexible locomotion system for robotic inspection in tubular confined spaces. Based on this concept, we designed a soft actuation mechanism that is able to adapt to different lumen sizes through a dedicated mechanical inflation mechanism, using the deformation of integrated elastic membranes. We implemented a proof-of-concept prototype which we validated experimentally. Experimental results showed that the locomotion mechanism is able to traverse through tubes of different diameter sizes, shapes and payloads with an average of (70%) locomotion efficiency across all testing conditions at diameter ratios spanning from (0.7) to (1.5). Moreover, the mechanical inflation mechanism was able to produce ( $\approx 13$  N) of normal (holding) force which is capable of supporting ( $\approx 5.8$  Kg) of payload including the weight of the robot, in a vertical tube configuration, assuming a static coefficient of friction ( $\mu_s = 0.5$ ) between the flexible sliders and the tubular environment.

In the future, we intend to advance the robot capabilities by creating a self-contained motorized version, featuring a flexible spine to enable steering and improved adaptation to varying pipeline curvatures. Subsequently, we aim to subject the robot to more realistic pipeline settings, evaluating its performance in real-world scenarios. Additionally, we plan to assess the robot's locomotion efficiency in soft pipeline environments, with a specific focus on exploring its viability for medical applications.

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# CHAPTER 7

## OSCAR: An Ovipositor-inspired Self-Propelling Capsule Robot for Colonoscopy

Under review as: M. A. Atalla, A. S. Sekar, R. van Starckenburg, D. J. Jager, A. Sakes, M. Wiertlewski and P. Breedveld, '*OSCAR: An Ovipositor-Inspired Self-Propelling Capsule Robot for Colonoscopy*' [1].



## Abstract

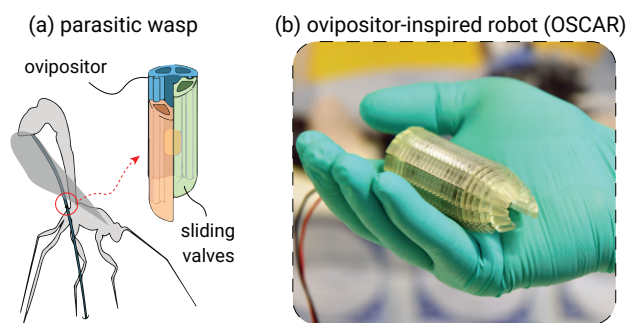
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Self-propelling capsule robots offer a route toward safer and more comfortable colonoscopy by eliminating the shaft looping associated with conventional push-driven insertion. However, generating reliable propulsion in the slippery, viscoelastic environment of the colon remains challenging. Following the self-propelling strategy developed in Chapter 6, this chapter presents OSCAR, an ovipositor-inspired self-propelling capsule robot for colonoscopy that uses coordinated multi-contact motion to tune system-level friction anisotropy and thereby generate net forward thrust. OSCAR mechanically encodes this motion pattern through a spring-loaded cam system that drives twelve circumferential sliders in a coordinated, phase-shifted sequence. Force characterization experiments in *ex vivo* porcine colon revealed a mean steady-state traction force of approximately 0.85 N, while locomotion experiments demonstrated an average speed of 3.08 mm/s, sufficient to match the cecal intubation times of conventional colonoscopy. The experiments further showed that thrust scales linearly with phase asymmetry while remaining largely independent of actuation speed. To explain these observations, the chapter develops an analytical modeling framework based on Kelvin–Voigt stick–slip interactions, linking phase asymmetry to thrust magnitude and slider-reversal synchronization to thrust stability. Beyond OSCAR, this framework provides a general basis for modeling related multi-contact self-propelling capsule systems. This chapter concludes Part II by establishing motion coordination as a predictive and controllable strategy for tuning system-level friction anisotropy to ‘propel’.

## 7.1. Introduction

Colorectal cancer (CRC) remains a major global health burden, ranking as the third most diagnosed malignancy and the second leading cause of cancer-related death worldwide, with recent estimates reporting 1.9 million new cases and nearly 1 million deaths each year [2, 3]. The most effective way to reduce this mortality is timely screening with colonoscopy. As the only modality that enables detection and removal of precancerous adenomas in a single session, colonoscopy reduces CRC mortality by 68% [4–6]. Supported by this proven clinical value, more than 19 million procedures are performed annually in the United States and approximately 50 million procedures are performed each year worldwide. [7, 8]. Yet, despite its clinical significance, conventional colonoscopy faces fundamental mechanical limitations that increase procedural difficulty for clinicians and compromise clinical outcomes.

Conventional colonoscopy relies on the manual advancement of a flexible endoscope through a tortuous and highly compliant anatomy [9]. Variability in colon anatomy across the population, ranging from differences in length and diameter to mobility, creates unpredictable boundary conditions that complicate instrument navigation and results in a steep learning curve for novice clinicians [10–12]. Pushing the colonoscope from the proximal end frequently causes its shaft to loop, especially in mobile segments such as the sigmoid, due to the dynamic interaction between the flexible shaft and the highly deformable and stretchable nature of the colon [13, 14]. This looping is the main reason behind patient pain and discomfort [15] as it subjects the colon walls to excessive forces, resulting in patient pain and forcing operators into successive motion cycles of pulling, twisting, and pushing the colonoscope forward [13] to resolve the loop and to have a path forward again. Furthermore, the control handle of conventional colonoscopes is difficult to use. Its design requires the user to carry the scope while operating two concentric knobs simultaneously to control the scope tip configuration, which is non-ergonomic and demands the operator to mentally map the rotations of two knobs to 3D tip movements while exerting insertion force, increasing cognitive load [16] and contributing to a high prevalence of musculoskeletal injuries among endoscopists [17, 18]. In challenging cases, these mechanical and ergonomic limitations increase the risk of failed cecal intubation, necessitating deeper sedation or alternative imaging modalities [19]. These limitations underscore the need for new technologies that reduce pain, improve patient comfort, improve controls, and reduce operator stress.



**Figure 7.1. Ovipositor-Inspired Capsule Robot for Colonoscopy (OSCAR).** (a) Parasitic wasps use reciprocating sliding valves that alternately anchor and release to transport eggs. (b) OSCAR translates this principle into a compact capsule with reciprocating sliders that generate self-propelling motion.

Robotic approaches aim to overcome the mechanical limitations of conventional colonoscopy by relocating actuation from the proximal handle to the distal tip. Instead of pushing the instrument and inducing loop formation, robotic systems seek to pull or guide the device through the lumen, thereby reducing both forces and stretch of the colon wall. These concepts have been explored through two broad strategies: external and internal actuation. External actuation uses electromagnetic coils or permanent magnets integrated into a robot arm to manipulate and steer a capsule or flexible endoscope equipped with an internal magnetic element [20–24]. Although this approach successfully eliminates looping and enables nearly painless navigation, it demands large, costly infrastructure and suffers from force limitations in patients with a high body mass index [24, 25]. These constraints have motivated a shift toward internal actuation approaches capable of self-propelling through on-board mechanisms, with the goal of achieving the safety benefits of distal actuation without the logistical requirements of magnetic platforms.

The actuation architecture of self-propelling endoscopes follow either a wireless or a tethered design. Wireless capsule endoscopy offers a non-invasive diagnostic experience, but remains fundamentally limited by its passive propulsion, restricted on-board energy (battery life), and lack of biopsy/therapeutic capabilities [25–27]. Consequently, achieving the functional capabilities of a conventional colonoscope still requires a physical connection. This need has driven the development of self-propelling tethered robotic capsules [28], which incorporate mechanisms onboard to generate propulsion while pulling a lightweight, flexible tether supplying power, fluids, and high-quality data transmission. By combining the mobility of a capsule with the functional

robustness of a tether, such devices represent a promising pathway toward painless yet fully therapeutic colonoscopy.

Several self-propelling architectures have been proposed in the literature to replace the push-driven colonoscope, including inchworm and crawler robots, track- and legged-based devices, and self-growing “vine” robots. Inchworm-type systems advance by sequentially anchoring and releasing their moving segments, typically via vacuum/suction or inflatable balloons [28–32], and have demonstrated acceptable diagnostic performance in human trials [33–35]. However, their inherently discontinuous clamp–extend gait leads to a substantially long cecal intubation time of ( $51 \pm 22.5$  min) on average [35], compared to (5 – 15 min) in routine practice and training benchmarks of conventional colonoscopy [11, 36], limiting their adoption beyond niche indications.

Alternatively, friction-driven robots that employ legs, paddles, or tracks can achieve higher speeds and more stable locomotion [37–41]. However, in the colon, their performance is fundamentally limited by the low effective friction coefficient of the mucosa: generating sufficient traction often requires elevated normal forces, which increases the risk of mucosal trauma. In addition, tether drag, particularly for pneumatically-driven capsules, and miniaturization required for capsule designs remain major practical challenges [42]. More recent soft self-growing colonoscope prototypes address the friction problem by growing from the tip [43–46], and thus minimizing forces on the wall compared to standard colonoscopy [47]. However, these systems still face unresolved challenges in retraction, reliable steering, integration of tools, and dependence on bulky pneumatic infrastructure, keeping them in the prototype stage.

As a result, these limitations suggest the need for a self-propelling solution that simultaneously offers compactness, high-speed, yet robust propulsion at low normal loads, robust towing of a functional tether, and a safety profile suitable for routine screening.

Addressing this unmet need, we present OSCAR: an ovipositor-inspired capsule robot for colonoscopy (Fig.7.1), a self-propelling capsule robot that takes inspiration from the way parasitic wasps drive their slender ovipositor needles through dense substrates, translating this biological strategy into a locomotion design concept suitable for the colon. This work makes four main contributions: First, we introduce a self-propelling locomotion principle capable of generating a continuous, tissue-safe thrust without relying on large normal forces. Second, we present a novel design

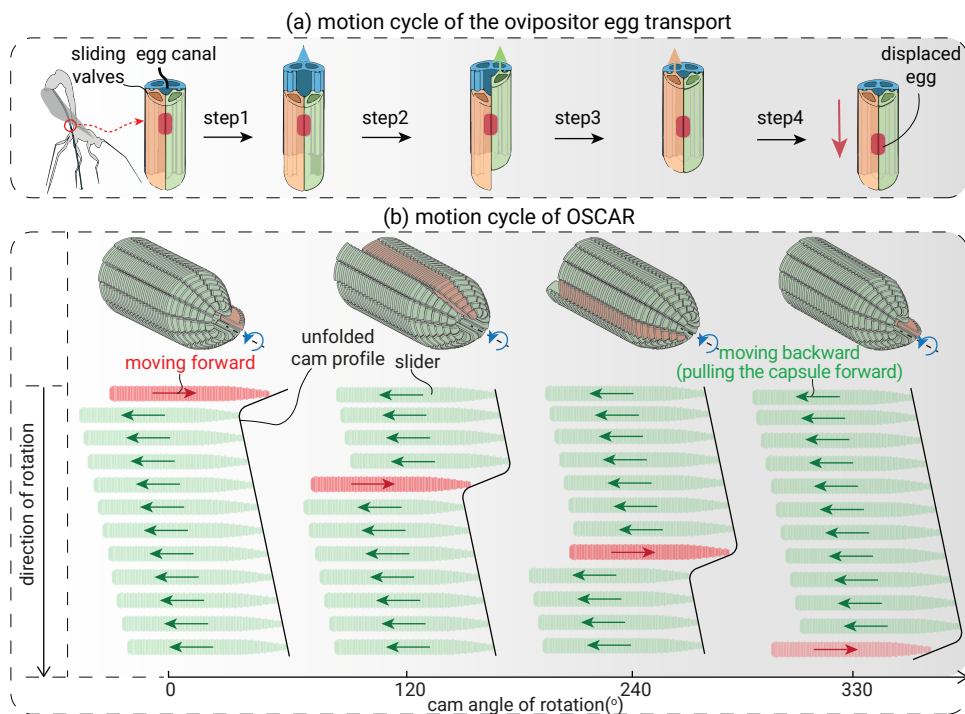
that realizes this self-propelling principle, which we implement in a physical prototype. Third, we develop an analytical modeling framework to mathematically describe the interaction between the capsule and the tissue wall, which we use to understand OSCAR's thrust generation mechanism, and to uncover how the design parameters, operating conditions and the capsule-tissue interaction properties (viscoelasticity and friction) affect the thrust, and to explore the design space beyond the presented prototype. Finally, we validate the locomotion principle and the mathematical model through systematic force-characterization and locomotion experiments in *ex vivo* porcine colon.

In the remainder of this chapter, Section 7.2 examines the biological inspiration in detail, outlining the locomotion principle of the wasp ovipositor. Section 7.3 presents the conceptual design and implementation of OSCAR, from the bio-inspired mechanism design to the physical prototype. In Section 7.4, we develop a mathematical model that captures OSCAR's kinematics and its interaction with the colonic wall, which we then use in Section 7.5 to explore the design space through simulation. Section 7.6 describes the force characterization and locomotion validation experiments, and Section 7.7 presents and discusses their results, implications and limitations. Finally, Section 7.9 summarizes the main findings and directions for future work.

## 7.2. Bio-inspiration from Wasp Ovipositor

Biological systems have evolved efficient strategies for locomotion through confined substrates, exemplified by the ovipositor of parasitic wasps. The ovipositor is an exceptionally slender, flexible organ that allows the wasp to penetrate substrates and deliver eggs deep inside host tissues (Fig. 7.2(a)). Its functionality relies on a set of three longitudinal valves, one dorsal and two ventral, arranged in a tongue-and-groove configuration that mechanically locks them together radially while allowing independent sliding along their length [48–52].

A key feature of this biological system is its ability to achieve controlled transport within a narrow tubular conduit without relying on large normal forces. Current theories suggest that the wasp induces frictional anisotropy between the valves and the egg to create net transport motion by coordinating the motion sequence of its valves [49, 52]. During each cycle, one valve retracts while the other two remain stationary to maintain higher frictional grip on the egg, effectively locking it in place. The motion then alternates sequentially between sliders until all sliders are in retraction



**Figure 7.2. Bio-inspiration and conceptual locomotion principle of OSCAR.** (a) Schematic representation of a parasitic wasp transporting eggs through coordinated stick–slip interactions generated by three sliding valves in their ovipositor. During each cycle, one valve retracts while the other two remain stationary, holding the egg in place because of the higher friction. When all valves reach their retract position, they advance together, producing a net forward displacement of the egg. Repeating this sequence enables controlled transport of the egg through the lumen till its deposition. (b) OSCAR translates and extends this principle into a capsule architecture using circumferential sliders driven by a rotating cam profile. Unlike the biological mechanism, where one valve moves at a time while the rest remain stationary, OSCAR drives a slider forward while driving the remaining sliders in the backward during each stroke. This continuous, phase-shifted reciprocation can yield a steady and continuous steadier average forward thrust suitable for self-propelling locomotion in colonoscopy.

position, after which all three slide forward together, advancing the egg by a stroke length. This coordinated stick–slip sequence produces reliable forward transport using only internal reciprocating motions. The wasp ovipositor, therefore, exemplifies a low-force strategy for propulsion in confined spaces, well suited for self-propelling endoscopic systems.

The transport concept of wasp ovipositors has inspired several medical and robotic innovations, including self-propelling needles that advance through soft tissue using interlocking sliding segments [53, 54], friction-based tissue-transport mechanisms that move samples through narrow channels [55], and soft robots for pipeline inspec-

tion [56]. Collectively, these examples highlight the broad applicability of ovipositor-inspired locomotion and its potential to address propulsion challenges in colonoscopy.

### 7.3. Design & Implementation of OSCAR

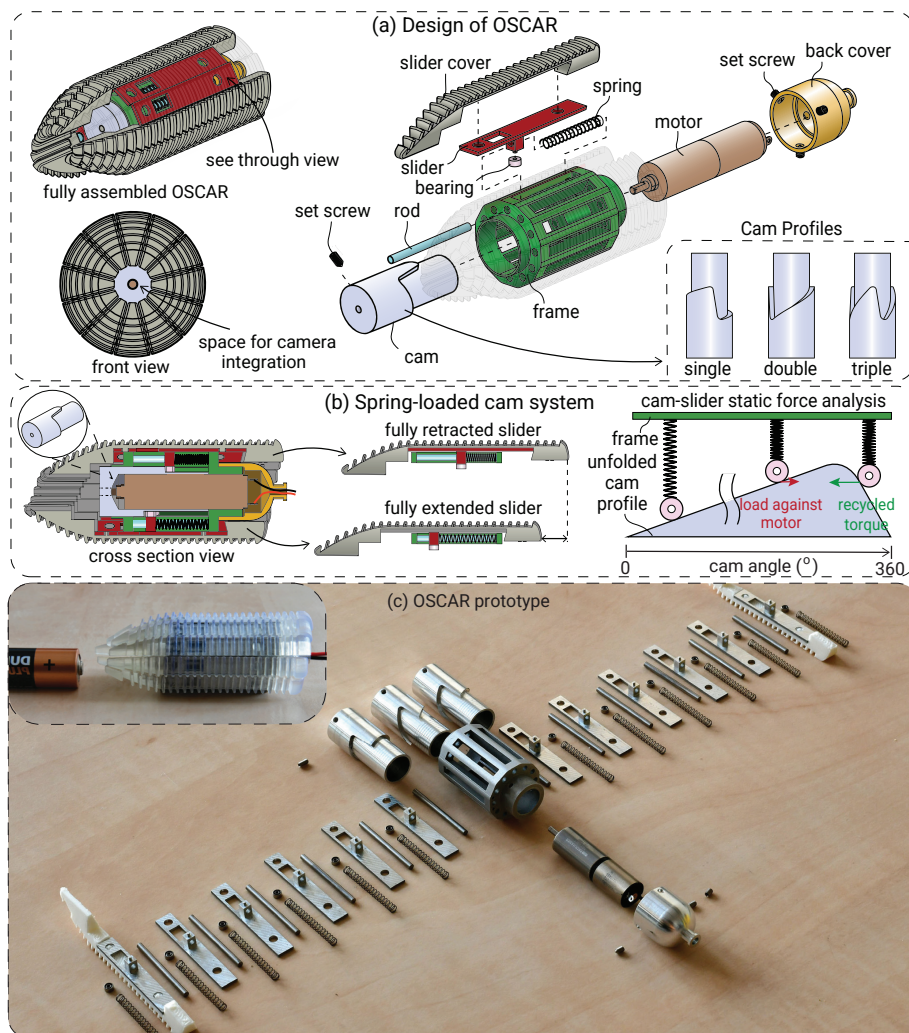
Inspired by the ovipositor's strategy of generating locomotion through coordinated sliding of its interlocked valves, OSCAR translates this biological principle into a capsule-scale propulsion concept suitable for the colon. The core idea is to distribute multiple longitudinal sliders evenly around the capsule body and drive their reciprocating motion using a rotating cam, as shown in Fig. 7.2(b). As the cam turns, each slider follows a prescribed phase-shifted trajectory, which sequences its forward and backward strokes relative to the others. This cam-driven coordinated motion enables a single drive input for all sliders and allows for steady thrust generation by simultaneously moving all sliders, in contrast to the discrete stepwise gait of the ovipositor.

The net forward thrust requires an asymmetry in the wall interaction such that, at any instant, more sliders retract than advance ( $n_{ret} > n_{adv}$ ), analogous to the ovipositor where one valve advances while the others hold position. This phase asymmetry creates an anisotropic frictional condition at the capsule–wall interface: the retracting sliders collectively generate a larger friction force that pushes the capsule forward, than the opposing friction produced by the advancing sliders, such that  $\sum F_{ret} > \sum F_{adv}$ , with  $\sum F_{ret} = n_{ret}F_f$  and  $\sum F_{adv} = n_{adv}F_f$ , where  $F_f$  is the friction force of a single slider. The resulting friction imbalance yields a nonzero net forward thrust, enabling self-propulsion in the same way that coordinated valve sliding transports eggs along the ovipositor. As such, achieving this force imbalance requires a minimum of three sliders, such as in the case of the wasp ovipositor.

Beyond nature's minimalist design, increasing the number of sliders is desirable, as it can further bias the cycle toward more retracting than advancing sliders and thereby strengthen the friction anisotropy and its resulting force imbalance, potentially yielding higher net thrust. Consequently, maximizing the number of sliders of the capsule becomes an important design consideration.

Overall, this conceptual design establishes a sliding system architecture that captures the essence of ovipositor-inspired locomotion while adapting to the colonoscopy application quest for steady and stable propulsion.





**Figure 7.3. Design and prototype of the OSCAR capsule robot.** (a) The design of OSCAR consists of 12 spring-loaded sliders distributed evenly around the circumference of the capsule body and driven by a central cam mechanism to realize the bio-inspired motion cycle described in the working principle. Each slider is guided by a cylindrical rod to ensure purely axial motion and is covered by a grooved slider cap to enhance frictional contact with the mucosa. The cam is mounted concentrically on a motor and enclosed by a back cover. Different cam profiles with varying numbers of motion reversals (jumps) were designed and fabricated to study the influence of motion pattern on thrust. (b) The spring-loaded cam system enables the use of steep cam angles while maintaining continuous contact between the sliders and the cam profile. This configuration allows maximum number of sliders to remain engaged in retraction, maximizing the potential net thrust. During the advance (fall) phase, the spring-loaded force of the advancing slider acting on the cam profile recycles the stored spring energy by generating a torque that assists the motor, thereby reducing the required actuation load. (c) Assembled and disassembled prototype of OSCAR, showing all key components of the actuation and transmission mechanisms.



### 7.3.1. Design Requirements

The design of OSCAR is guided by a set of anatomical and mechanical requirements that the capsule must satisfy to function reliably inside the human colon. The colon diameter varies substantially along its length, ranging from approximately 30 mm in the sigmoid colon to over 80 mm in the cecum [57]. To ensure safe passage through bends and local constrictions, the capsule diameter must therefore remain below the minimum diameter of the sigmoid colon ( $< 30$  mm).

In addition to the size requirement, the system must meet practical procedure-time targets. To achieve a cecal intubation time comparable to conventional colonoscopy (5 – 15 min) [11, 36], the capsule should traverse a colon of  $\sim 1.5$  m median length (and potentially  $> 1.8$  m) [58], which implies an average forward speed on the order of  $\sim 2$ – $6$  mm/s.

The propulsion efficiency is further constrained by the tribological properties of the colon. The mucosa is coated with a lubricating mucus layer, whose coefficient of friction with the capsule can range from very low  $\mu \approx 0.03$  to relatively high  $\mu \approx 0.5$ , depending on the material and design of the capsule sliders, the lubricity of the colon wall, and the normal load between the capsule and the colon wall [59–61]. To achieve locomotion, the capsule must generate net thrust force that overcomes the static friction with the wall, while also carrying its own load and counteracting any additional resistance sources such as tether drag. For similar capsules, the net thrust force reported is typically around 1.0 N [61].

These performance targets must remain within strict safety boundaries. Local contact shear forces should stay within safe limits to avoid superficial mucosal trauma and, critically, prevent damage beyond the mucosal layer [60]. The propulsion mechanism must therefore balance traction generation with tissue protection, producing consistent forward force while minimizing local pressure concentrations.

Together, these geometric, kinematic, and tribological requirements define the design space for OSCAR.

### 7.3.2. Detailed Design of OSCAR

Building on our conceptual design and taking into account the design requirements, we strived for a capsule that complies with the size constraint ( $< 30$  mm diameter) while still maximizing the number of sliders to increase the maximum attainable

thrust, as implied by our design concept. We therefore selected a capsule diameter of 28 mm to leave a safety margin below the size limit. Within this diameter, we were able to fit a maximum of twelve sliders while keeping the design realistic to fabricate and assemble; beyond this number, the feature sizes and tolerances become hard to manufacture. The sliders are distributed evenly around the circumference of the frame and are constrained to translate axially along dedicated grooves machined into the main frame, as illustrated in Fig. 7.3(a).

Each slider carries a cover to interface with the colon wall. The cover features a slotted surface design to enhance frictional grip against the intestinal wall by trapping mucus and pushing any fluid at the interface into the slots, and leveraging local deformation of the colon wall between the slots to achieve better grip [60]. The slot ridges are symmetric (i.e., not inclined in either direction) to ensure that sliding friction is symmetric. The cover is rounded (oval) at both ends to ensure smooth interaction at the capsule boundaries.

To implement the alternating retract-advance pattern inspired by the ovipositor mechanism, the cam profile actuating the sliders was designed to maximize the instantaneous imbalance between retracting and advancing sliders, thereby strengthening the motion asymmetry and the potential net thrust. In addition, we strived to provide sufficient stroke length to overcome the elastic wall reaction and break the static friction, which is necessary to propel forward. We defined the slider stroke length to be (10 mm).

Producing a cam profile that achieves this stroke length while maximizing the ratio of retracting to advancing sliders requires steep motion reversal angles. For example, enabling a maximum of eleven sliders to retract while only one advances implies that the advance phase occupies 1/12 of the cam cycle, requiring reversal angles  $> 80^\circ$  to push forward the slider in the advance phase. Under these conditions, traditional grooved cam–follower designs are not feasible: such steep reversals would require very high actuation torque to force the followers through the corners, or otherwise the sliders would be mechanically-locked and the motion would stall. To overcome this challenge, we designed a single-sided cam profile (rather than a closed groove), where the sliders are preloaded against the cam surface with springs to maintain contact as the cam rotates. This spring-loaded cam–slider design prevents mechanical locking and simplifies the torque requirements.

A positive side effect of this design is that the slider(s) in the advance phase can recycle the spring energy accumulated during retraction by pushing against the cam

wall, generating an assisting torque that supports the motor and partially relieves it from the load, as illustrated in Fig. 7.3(b). In this configuration, the steep reversal angle becomes beneficial because they increase the force component that contributes to this assisting torque. For experimental validation, three cam profiles were developed with one, two, and three motion reversals (“jumps”) per revolution, as shown in Fig. 7.3(a), enabling systematic variation of the advance and retract timing. Each cam was machined with the motion profile around its circumference while being hollow to house the driving motor, reducing the axial length of OSCAR and making the device more compact, as shown in Fig. 3(b). We selected a pre-loading spring of 5 N/mm stiffness to ensure reliable tracking of the cam profile under varying tissue stiffness and surface lubricity conditions.

The final OSCAR design measures 28 mm in diameter and 60 mm in length, with a slider stroke length of 10 mm and a total assembled mass of approximately 50 g, as indicated in Table 7.1. The overall architecture is modular, allowing rapid cam exchange and straightforward assembly while preserving space at the anterior end of the capsule for potential imaging integration.

## 7

### 7.3.3. Prototype Development

The prototype of OSCAR was developed to validate its working principle and assess its locomotion performance. Structural components were fabricated using milling and electric discharge machining (EDM) from aluminium 7075–T5, which we selected due to its high strength-to-weight ratio and dimensional stability under load. To accommodate the intricate features of slider covers, these covers were additively manufactured using stereolithography (SLA) and subsequently bonded to the machined sliders. Standard off-the-shelf components—including compression springs, guide rods, and miniature bearings—were integrated to streamline assembly and facilitate future iterations. The prototype was driven by a 12-V DC motor with an integrated gearbox to deliver the torque required to operate the capsule (1016M012S, FAULHABER, Schönaich, Germany). Figure 7.3(c) shows both the assembled and disassembled states of the prototype, highlighting the compact arrangement and the mechanical interfaces between the frame, cam, motor, sliders, and springs.

**Table 7.1.** Design parameters of the OSCAR prototype.

| Design parameter | description      | Value |
|------------------|------------------|-------|
| $D_{capsule}$    | Capsule diameter | 28 mm |
| $L_{capsule}$    | Capsule length   | 60 mm |
| $h_{stroke}$     | Slider stroke    | 10 mm |
| $m$              | Mass             | 50 g  |

## 7.4. Modeling of OSCAR

In order to understand the thrust generation mechanism of OSCAR and its contributing factors, we built an analytical model that captures the kinematics of a single slider and its force interaction with the wall, which we then extended to the full capsule thrust by superimposing the phase-shifted contributions of the  $n$  sliders.

### 7.4.1. Single Slider Kinematics

Consider a single slider from the actuation system of OSCAR, spring-loaded against the cam as illustrated in Fig. 7.4(a). We assume that the spring preload maintains continuous contact between the slider and the cam surface, such that separation or backlash between the slider and the cam profile does not occur during the cam cycle. Under this assumption, the slider is kinematically constrained to the cam which means that its axial displacement is fully prescribed by the cam profile and does not depend on inertia or spring dynamics. This assumption holds when the preload force exceeds the resisting forces due to slider inertia during cam actuation and due to slider-wall interaction.

The cam profile is described by a piecewise lift function  $h(\theta)$ , which gives the imposed slider displacement as a function of the cam angle  $\theta$ . In all cases, the cam law is expressed as a continuous piecewise function composed of linear and circular segments that represent the rise, fall, and dwell portions of the cam (Fig. 7.4). This can be written in general form as follows:

$$h(\theta) = \begin{cases} h_1(\theta), & \theta_0 \leq \theta < \theta_1, \\ h_2(\theta), & \theta_1 \leq \theta < \theta_2, \\ h_3(\theta), & \theta_2 \leq \theta < \theta_3, \\ \dots & \text{until } 2\pi, \end{cases} \quad (7.1)$$

where  $h_i(\theta)$  denotes the  $i$ -th segment of the cam law defined on the angular interval  $[\theta_{i-1}, \theta_i)$ , with  $0 \leq \theta_0 < \theta_1 < \theta_2 < \dots \leq 2\pi$ . Each  $h_i$  corresponds to either a circular arc or a linear segment as defined by the cam design.

Assuming that the cam rotates with a constant rotation speed  $\omega$ , the cam angular displacement  $\theta(t)$  can be described as follows:

$$\theta(t) = \omega t, \quad \theta \in [0, 2\pi]. \quad (7.2)$$

Since the slider is kinematically constrained to the cam, we can find the slider displacement corresponding to the cam rotation by substituting the cam angle expression Eq. 7.2 into the cam profile lift function Eq. 7.1, as follows:

$$x_{slider}(t) = h(\theta(t)) \quad (7.3)$$

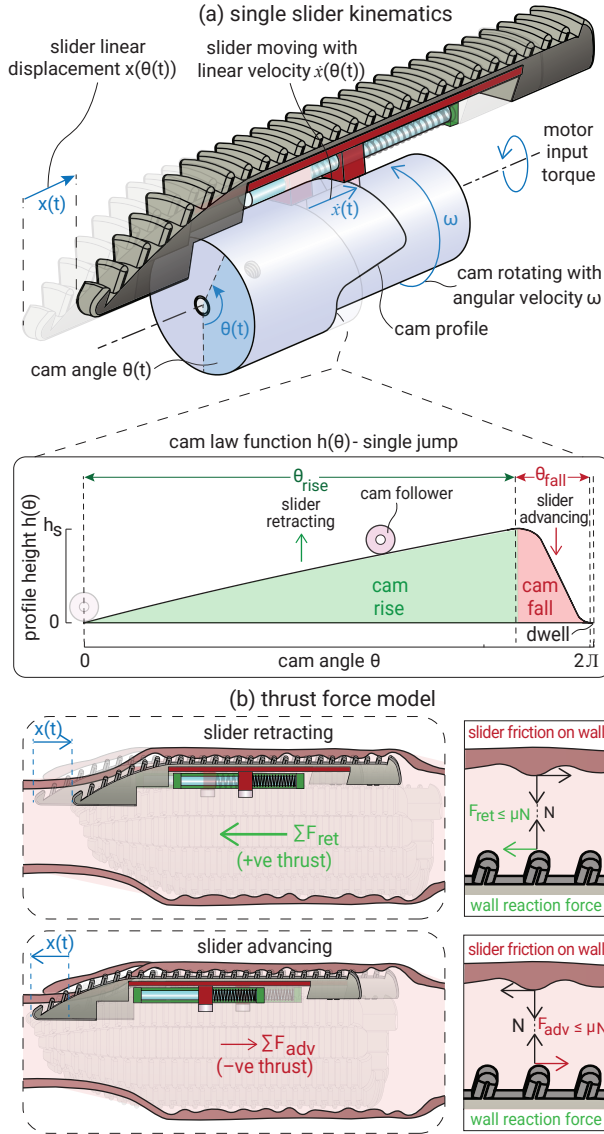
By applying the chain rule and differentiating the slider displacement  $x_{slider}(t)$  in time, we can find the corresponding slider velocity:

$$\dot{x}_{slider}(t) = \frac{dh}{d\theta} \frac{d\theta(t)}{dt} = \omega \frac{dh}{d\theta}(\theta(t)) \quad (7.4)$$

We can see from Eqs. (7.2)–(7.4) that assuming a constant cam rotation speed  $\omega$  results in time-invariant kinematics, which means that the slider motion depends solely on the cam angle  $\theta$  and not explicitly on time. Consequently, for every identical angular position of the cam, the slider possesses the same displacement, velocity, and acceleration states.

#### 7.4.2. Single Slider Instantaneous Force

When the capsule interfaces with the colon wall through the sliders, the wall and the capsule press against each other with a normal load  $N$ , which can originate, for example, from the deflated baseline configuration of the colon or from intra-abdominal pressure acting on the wall. Each individual slider is therefore subjected to this normal force. We define forward capsule thrust as positive. When the slider retracts, meaning the slider moves backward relative to the capsule, the wall pushes forward



**Figure 7.4. Single-slider kinematics and thrust force generation.** (a) Kinematic description of OSCAR's slider actuation. A cam rotating with angular velocity  $\omega$  drives each slider through a prescribed linear trajectory  $x(t)$ . The cam law for the single-jump profile consists of a slow rise phase (slider retraction), a steep fall phase (slider advance), and a short dwell. The interval asymmetry between rise and fall sets the duty-cycle imbalance that underpins net thrust generation. (b) Thrust force model based on frictional interaction with the colonic wall. During retraction (top), the slider moves backward relative to the capsule, generating positive thrust as the wall traction opposes the slider motion. During advance (bottom), the slider moves forward, producing negative thrust as the wall reaction reverses direction. Superposition of these phase-shifted contributions of all sliders yields the average net thrust force.

on the capsule with a reaction force  $F_{ret}$ . Conversely, when the slider advances, meaning the slider moves forward relative to the capsule, the wall reaction is directed backward and is denoted  $F_{adv}$ . Both forces,  $F_{ret}$  and  $F_{adv}$ , are limited by the available friction at the interface.

Given our slotted slider design which helps pushing mucus and interface fluids to the slots while maintaining contact with the ridges, we assume that frictional interaction between the capsule and the wall follows the Coulomb friction law described by the coefficient of friction  $\mu$  between the slider and the colon wall and the normal force at the interface  $N$ , in line with prior capsule–tissue models [29, 62–64]. Accordingly, the retraction and advance forces are bound by the friction limit at the interface, as follows:

$$|F_{adv}|, |F_{ret}| \leq \mu N, \quad (7.5)$$

The instantaneous slider traction force ( $F_{slider}$ ) alternates between the retraction and advance forces depending on its instantaneous velocity (i.e. motion direction, retracting or advancing) during the cycle, which can be expressed mathematically as follows:

$$F_{slider}(\theta) = \begin{cases} \leq +\mu N, & \dot{x}_{slider} < 0 \quad (\text{retraction}), \\ \geq -\mu N, & \dot{x}_{slider} > 0 \quad (\text{advance}), \\ 0, & \dot{x}_{slider} = 0 \quad (\text{dwell}). \end{cases} \quad (7.6)$$

Depending on the sliding state, sticking or slipping, the instantaneous slider traction force is bounded by  $|F_{slider}| \leq \mu N$ . When the slider sticks, i.e. it does not move relative to the colon wall, the slider traction force remains within the bounds of the friction limit  $|F_{slider}| \leq \mu_s N$ , governed by the static coefficient of friction,  $\mu_s$ . When the slider slips, meaning that it moves relative to the colon wall, the slider traction force saturates at the friction limit  $|F_{slider}| = \mu_k N$ , which is governed by the kinetic coefficient of friction  $\mu_k$  where ( $\mu_k < \mu_s$ ). These two sliding conditions that define the friction coefficient can be expressed mathematically as follows:

$$\mu = \begin{cases} \mu_s, & \dot{x}_{slider}^{wall} = 0 \quad (\text{stick}), \\ \mu_k, & |\dot{x}_{slider}^{wall}| > 0 \quad (\text{slip}). \end{cases} \quad (7.7)$$

Where  $\dot{x}_{slider}^{wall}$  is the velocity of the slider relative to the colon wall.

The idealized Coulomb friction law in Eq.7.6 assumes that the contact force switches instantly between positive and negative at motion reversals. However, in reality, the compliance of the colon wall and the micro-slip at the slider-wall interface lead to gradual force transitions instead of abrupt reversals. This behavior can be captured by introducing a smoothing function  $f(\lambda)$  to the instantaneous slider force  $F_{slider}$ , where  $\lambda$  is a smoothing parameter that controls how smooth the transition is. This function matters only in the fractional portion of the cycle over which the slider reverses its direction of motion. The instantaneous slider force can therefore be expressed in a generalized form as:

$$F_{slider}(\theta) = \mu N f(\lambda, \theta) \quad (7.8)$$

where  $f(\lambda)$  varies between +1 and -1 over the transition period. Assuming  $\lambda \in [0, 1]$ , the limit  $\lambda = 0$  recovers the instant force reversal, while larger values of  $\lambda$  correspond to smoother force reversals. This parameter  $\lambda$  can be regarded as an empirical parameter that captures the combined effect of contact relaxation, material compliance, and lubrication on the transition smoothness.

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### 7.4.3. Single Slider Cycle-Averaged Force

To describe the periodic force generated by a slider in a cam cycle, we define duty fractions  $d_{adv}$  and  $d_{ret}$ , which represent the spatial fractions of the cycle spent in advance and retraction, respectively. These fractions are determined by the cam profile (including any dwells) and satisfy ( $0 \leq d_{adv}, d_{ret} \leq 1$ ) with ( $d_{adv} + d_{ret} \leq 1$ ). The cycle-averaged thrust of a single slider can then be expressed as

$$\bar{F}_{slider} = F_{ret} d_{ret} - F_{adv} d_{adv} - F_{loss}, \quad (7.9)$$

where  $F_{loss}$  represents internal mechanical losses such as friction of the slider guide.



### 7.4.4. Multi-Slider Instantaneous Force Superposition

Assuming the capsule employs  $n$  identical sliders that are uniformly phase-shifted by  $2\pi/n$  around the cam profile. The instantaneous net thrust of the capsule is the superposition of all slider contributions:

$$F_{\text{capsule}}(\theta) = \sum_{i=1}^n F_{\text{slider}}(\theta) = \sum_{i=1}^n F_{\text{slider}}\left(\theta - \frac{2\pi(i-1)}{n}\right). \quad (7.10)$$

Because the sliders are evenly distributed in phase, at any cam angle a certain number will be in advance and the rest in retraction. The ratio between advancing and retracting sliders is determined by the duty fractions ( $d_{\text{adv}}$ ,  $d_{\text{ret}}$ ) introduced earlier. If a single slider spends a fraction  $d_{\text{adv}}$  of its cycle in advance and a fraction  $d_{\text{ret}}$  in retraction, then approximately the same fractions of the  $n$  sliders will be in those phases at any instant, which can be expressed as follows:

$$n_{\text{adv}}(\theta) \approx n d_{\text{adv}}, \quad n_{\text{ret}}(\theta) \approx n d_{\text{ret}}. \quad (7.11)$$

With small deviations during short overlap intervals where transitions occur.

## 7

### 7.4.5. Steady-State Cycled-Averaged Thrust Force

To find the steady-state thrust of the robot, we compute the cycle-averaged force over one cam revolution by applying the averaging operator to the instantaneous capsule force expression in (7.10) as follows:

$$\bar{F}_{\text{capsule}} = \frac{1}{2\pi} \int_0^{2\pi} \sum_{i=1}^n F_{\text{slider}}\left(\theta - \frac{2\pi(i-1)}{n}\right) d\theta. \quad (7.12)$$

Because the sliders are identical and uniformly phase-shifted, each term contributes the same average force, yielding:

$$\bar{F}_{\text{capsule}} = n \bar{F}_{\text{slider}}, \quad (7.13)$$

where  $\bar{F}_{\text{slider}}$  is the cycle-averaged thrust of a single slider over one cam cycle. By substituting the single slider cycle-averaged force Eq.(7.9) into the n-sliders capsule

cycle-averaged thrust force expression, we find the following expression for the capsule cycle-averaged thrust force:

$$\bar{F}_{\text{capsule}} = n \left( F_{\text{ret}} d_{\text{ret}} - F_{\text{adv}} d_{\text{adv}} - F_{\text{loss}} \right) \quad (7.14)$$

This formula shows that the cycle-averaged thrust force, that OSCAR produces, is governed by the number of sliders  $n$ , the geometric design of the cam profile represented by its duty fractions  $d_{\text{ret}}$ ,  $d_{\text{adv}}$ , the available traction at the interface  $\mu N$  represented by  $F_{\text{ret}}$ ,  $F_{\text{adv}}$ , and the internal losses per slider  $F_{\text{loss}}$ .

#### 7.4.6. Non-Dimensional Analysis

To generalize the model and facilitate comparison between different cam geometries and operating conditions, we conducted a non-dimensional analysis in which we express the system variables in non-dimensional form. The characteristic quantities used for normalization are the cam angular period  $2\pi$ , the maximum cam lift  $H = \max(h)$ , and the traction force limit  $\mu N$ . The non-dimensional parameters are therefore defined as follows:

$$\tilde{\theta} = \frac{\theta}{2\pi}, \quad \tilde{h}(\tilde{\theta}) = \frac{h(\theta)}{H}, \quad \tilde{F}_{\text{slider}}(\tilde{\theta}) = \frac{F_{\text{slider}}(\theta)}{\mu N}.$$

Substituting into the expression for the instantaneous slider force from Eq. (7.6) and using the non-dimensional cam slope  $\frac{d\tilde{h}}{d\tilde{\theta}}$ , the normalized slider force becomes:

$$\tilde{F}_{\text{slider}}(\tilde{\theta}) = f(\lambda, \tilde{\theta}) \text{sign} \left[ -\frac{d\tilde{h}}{d\tilde{\theta}}(\tilde{\theta}) \right], \quad (7.15)$$

where  $f(\lambda, \tilde{\theta})$  is the previously introduced smoothing function that governs the transition between the forward and reverse friction limits. In the ideal case of saturating at maximum traction ( $\mu N$ ) at all time, i.e. in case of sliders slipping at all time, the smoothing effect vanishes ( $\lambda = 0$ ), and  $\tilde{F}_{\text{slider}}$  alternates between  $\pm 1$ . Increasing  $\lambda$  produces smoother transitions corresponding to compliant or lubricated contact conditions.

The non-dimensional cycle-averaged thrust of a single slider can then be written as

$$\bar{\tilde{F}}_{\text{slider}} = \frac{1}{2\pi} \int_0^{2\pi} \tilde{F}_{\text{slider}}(\tilde{\theta}) d\tilde{\theta} = (d_{\text{ret}} - d_{\text{adv}}) - \bar{\tilde{F}}_{\text{loss}}, \quad (7.16)$$

where  $\tilde{F}_{\text{loss}} = F_{\text{loss}}/\mu N$  is the normalized internal loss.

For a capsule employing  $n$  uniformly phase-shifted sliders, the instantaneous non-dimensional thrust is given by

$$\tilde{F}_{\text{capsule}}(\tilde{\theta}) = \frac{1}{n} \sum_{i=1}^n \tilde{F}_{\text{slider}}\left(\tilde{\theta} - \frac{i-1}{n}\right), \quad (7.17)$$

and the corresponding cycle-averaged non-dimensional thrust is:

$$\bar{\tilde{F}}_{\text{capsule}} = (d_{\text{ret}} - d_{\text{adv}}) - \tilde{F}_{\text{loss}}. \quad (7.18)$$

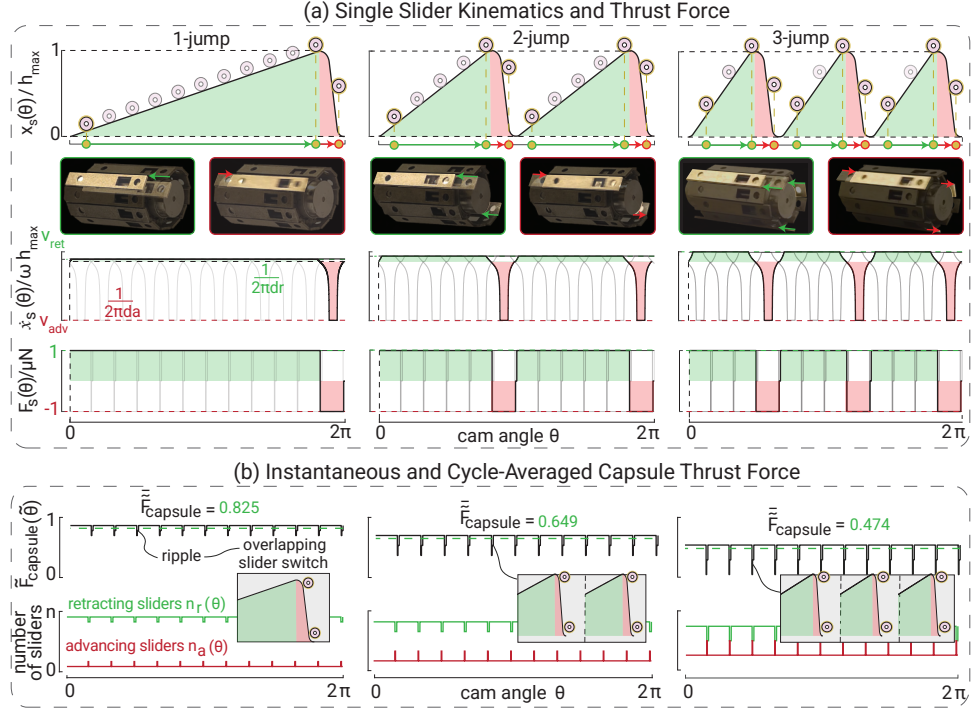
This expression emphasizes that the non-dimensional average thrust depends only on the duty asymmetry  $(d_{\text{ret}} - d_{\text{adv}})$  governed by the cam geometry and the loss term  $\tilde{F}_{\text{loss}}$  per slider.

## 7.5. Simulation of OSCAR

We used the analytical model in Section 7.4 to simulate OSCAR's kinematics and thrust under the ideal, friction-limited assumption that each slider's traction saturates at  $\mu N$ . The simulation study aimed, first, to clarify how the three implemented cam profiles translate into slider motion and capsule thrust, and second, to examine how the cam duty fractions  $(d_{\text{ret}}, d_{\text{adv}})$ , the jump count  $k$ , and the number of sliders  $n$  collectively shape both the instantaneous thrust profile and its cycle-average.

### 7.5.1. Kinematics and Thrust Force of Implemented Cam Designs

We first simulated the single-slider kinematics for our three implemented cam designs (with  $k = 1, 2, 3$ ) by representing each cam through its lift function  $h(\theta)$  and computing the slider displacement and velocity using Eqs. (7.2)–(7.6) (Fig. 7.5(a)). Due to the kinematic constraint between the slider and the cam profile, the slider displacement directly follows  $h(\theta)$ . Accordingly, the normalized displacement  $x_{\text{slider}}(\theta)/h_{\text{max}}$  (Fig. 7.5(a)) exhibits one, two, and three rise–return motion reversals over one revolution for  $k = 1, 2, 3$ , respectively. The corresponding normalized velocity profiles  $\dot{x}_s(\theta)/(\omega h_{\text{max}})$  indicate how rapidly each stroke is executed and when the transitions between advance and retract occur (zero crossings) over the full  $2\pi$  cycle. These transition angles define when a slider contributes forward versus



**Figure 7.5. Simulation of kinematics and thrust force of OSCAR.** (a) Single-slider displacement, velocity, and instantaneous traction force for 1-, 2-, and 3-jump cam geometries. Each cam produces distinct rise–return sequences corresponding to retracting (green) and advancing (red) phases, with normalized slider velocity given by  $\dot{x}_{\text{slider}}^*/(\omega h_{\max}) = \pm 1/(2\pi d_{r,a})$ . (b) Instantaneous and cycle-averaged capsule thrust force obtained by superposing the phase-shifted contributions of  $n = 12$  sliders. The upper plots show the total normalized thrust  $\bar{F}_{\text{capsule}}(\theta)$  and its mean value  $\bar{F}_{\text{capsule}}$ , while the lower plots show the instantaneous number of retracting  $n_r(\theta)$  and advancing  $n_a(\theta)$  sliders. Increasing the number of jumps compresses the individual stroke, increases switching frequency, and reduces the average thrust, while amplifying ripple due to increased overlapping between slider transitions. Refer to Supplementary Video.1 for a demonstration of the motion cycles of the three cam profiles.

backward traction (Fig. 7.5(a)). In our three cam designs, we kept the advance-phase angle constant (red in Fig. 7.5(a)), and therefore the peak advance (negative) velocity is identical across cams. Increasing  $k$  thus compresses only the retract interval (green in Fig. 7.5(a)), which increases the peak retract (positive) velocity.

Using these kinematics, we evaluated the normalized instantaneous traction of a single slider,  $\tilde{F}_{\text{slider}}(\theta)$ , from Eq. (7.15) (Fig. 7.5(a)). Under the friction-limited assumption (sliding always in slip), the traction saturates at  $\pm 1$  (i.e.,  $\pm \mu N$  after normalization), with the sign set by the motion direction. We then obtained the normalized capsule thrust  $\tilde{F}_{\text{capsule}}(\theta)$  by superimposing the phase-shifted contributions of all sliders

( $n = 12$ ) using Eq. (7.17), and computed the cycle-average  $\bar{\tilde{F}}_{\text{capsule}}$  using Eq. 7.18 (Fig. 7.5(b)).

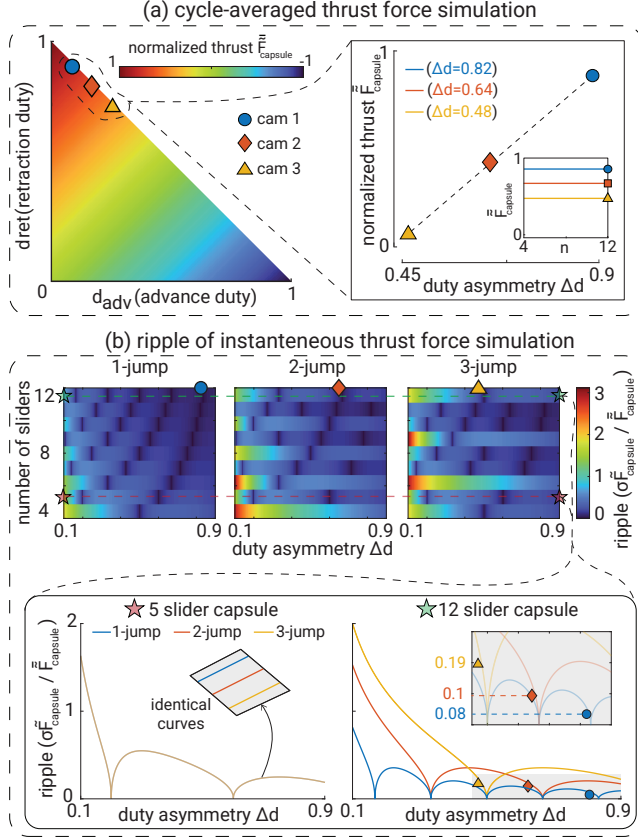
The computed instantaneous normalized capsule thrust  $\tilde{F}_{\text{capsule}}(\theta)$  shows two components (Fig. 7.5(b)): a mean value and a small periodic ripple around that mean. The mean values for the three cams reflect the cycle-averaged thrust  $\bar{\tilde{F}}_{\text{capsule}}$ , which is set by the duty asymmetry  $\Delta d = d_{\text{ret}} - d_{\text{adv}}$  of each profile (see Fig. 7.6(a)). The traces of the slider counts  $n_{\text{ret}}$  and  $n_{\text{adv}}$  over the cycle (Fig. 7.5(b)) reveal the origin of the ripple. Specifically, the ripple arises from brief count changes during motion reversals when the slider spacing does not align exactly with the angular lengths of the advance and retract intervals. In that case, one slider switches phase while the preceding slider has not yet left it, producing a transient change in  $n_{\text{ret}}$  and  $n_{\text{adv}}$ . This transient reduces temporarily the number of sliders contributing to forward thrust, resulting in the periodic ripple observed in  $\tilde{F}_{\text{capsule}}(\theta)$ .

## 7.5.2. Effect of Cam Duty Asymmetry, Jump Count, and Slider Count on Thrust

To study the effects of the cam duty fractions ( $d_{\text{ret}}, d_{\text{adv}}$ ), the cam jump count  $k$ , and the number of sliders  $n$ , we simulated the capsule across a range of duty fractions, jump counts, and slider counts, and quantified their effects on both the cycle-averaged thrust and the periodic ripple in the instantaneous thrust profile.

As shown in Fig. 7.6(a), the cycle-averaged normalized thrust  $\bar{\tilde{F}}_{\text{capsule}}$  is determined solely by the cam duty fractions ( $d_{\text{ret}}, d_{\text{adv}}$ ) and varies linearly with the duty asymmetry  $\Delta d = d_{\text{ret}} - d_{\text{adv}}$  independent of the jump count  $k$  and the number of sliders  $n$ . Because  $\Delta d$  sets the net imbalance between the retract and advance intervals over a cycle, it directly controls both thrust direction and magnitude: a positive duty asymmetry  $d_{\text{ret}} > d_{\text{adv}}$  ( $\Delta d > 0$ ) generates forward (positive) thrust, a negative duty asymmetry  $d_{\text{ret}} < d_{\text{adv}}$  ( $\Delta d < 0$ ) generates backward (negative) thrust, and a null duty asymmetry yields zero mean thrust (Fig. 7.6(a)). This highlights duty asymmetry as a potential handle for bi-directional propulsion.

To quantify the periodic ripple in the normalized instantaneous thrust  $\tilde{F}_{\text{capsule}}(\theta)$ , we define the ripple as the coefficient of variation over one cycle, i.e., the standard deviation normalized by the cycle-average,  $\sigma_{\tilde{F}_{\text{capsule}}} / \bar{\tilde{F}}_{\text{capsule}}$ . As explained earlier, this ripple arises from overlapping slider phase switches, which temporarily change the number of sliders in each phase (as in Fig. 7.5(b)). The duty asymmetry  $\Delta d$ , jump



**Figure 7.6. Simulation of cycle-averaged thrust and instantaneous thrust ripple of OSCAR.** (a) Duty-fractions map of the cycle-averaged normalized thrust  $\bar{F}_{\text{capsule}}$  over the cam duty fractions ( $d_{\text{ret}}, d_{\text{adv}}$ ). The map shows that the mean thrust is set by the imbalance between the retract and advance intervals:  $d_{\text{ret}} > d_{\text{adv}}$  yields forward (positive) thrust,  $d_{\text{ret}} < d_{\text{adv}}$  yields backward (negative) thrust, and  $d_{\text{ret}} = d_{\text{adv}}$  yields zero mean thrust. Markers indicate the three implemented cam profiles. The plot on the right shows that the normalized cycle-averaged thrust increases linearly with the duty asymmetry  $\Delta d = d_{\text{ret}} - d_{\text{adv}}$ , independent of number of sliders  $n$  (inset) and the number of cam motion reversals (jump count)  $k$ . (b) Ripple of the normalized instantaneous thrust  $\bar{F}_{\text{capsule}}(\theta)$ , quantified as the coefficient of variation  $\sigma \bar{F}_{\text{capsule}} / \bar{F}_{\text{capsule}}$ , for different slider counts  $n$  and jump counts  $k$ . Ripple arises from overlapping slider phase switches that temporarily change the numbers of sliders in retraction and advance. The heat maps and representative curves show that  $\Delta d$  shifts the phase boundaries within each of the  $k$  segments relative to the slider spacing  $2\pi/n$ , producing alternating low- and high-ripple regions as overlap is avoided or occurs. When  $n/k$  is an integer (e.g.,  $n = 12$ ), the transition pattern repeats across segments and ripple depends strongly on  $(\Delta d, k)$ ; when  $n/k$  is not an integer (e.g.,  $n = 5$ ), the transition pattern becomes asymmetric and thus the ripple becomes less sensitive to  $k$ .

count  $k$ , and slider count  $n$  therefore set when and how often these count changes occur.

Specifically,  $\Delta d$  sets the angular lengths of the retracting and advancing intervals within each of the  $k$  segments of the cycle, and therefore sets the phase boundaries (advance vs. retract) in each segment. As  $\Delta d$  is varied for a fixed  $k$ , these boundaries slide relative to the slider spacing  $2\pi/n$ , changing whether sliders switch phases sequentially (one-by-one) or with temporal overlap. As a result, the ripple (Fig. 7.6(b)) is governed by the interplay between the geometric slider spacing ( $2\pi/n$ ) and the cam timing ( $k, \Delta d$ ), producing alternating low- and high-ripple regions as  $\Delta d$  is swept for each  $k$ .

The role of  $k$  is to set how many rise–return sequences occur per revolution, and thus how many phase switches occur per cycle. When the  $n$  sliders are evenly distributed across the  $k$  segments (i.e.,  $n/k$  is an integer), the same transition pattern repeats from segment to segment, so overlap events recur at the same relative times and generate ripple that depends on the combination of  $\Delta d$  and  $k$ ; as reflected in the  $n = 12$  case across  $k = 1, 2, 3$  (bottom-right plot and heat maps in Fig. 7.6(b)). In contrast, when  $n/k$  is not an integer, sliders are not equally distributed across segments, so overlaps occur at different times in different segments and do not repeat consistently; consequently, the ripple becomes less sensitive to  $k$ , as illustrated by the identical curves for  $n = 5$  (bottom-left plot in Fig. 7.6(b)).

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### 7.5.3. Design Implications from Simulation

The simulations indicate that mean thrust and thrust smoothness are tuned by different parameters. The cycle-averaged thrust is set solely by the duty asymmetry  $\Delta d = d_{ret} - d_{adv}$ : it scales linearly with  $\Delta d$  and reverses sign with it, enabling bi-directional control through cam timing. In contrast, the jump count  $k$  and slider count  $n$  mainly affect thrust smoothness (ripple) by shaping (together with the duty asymmetry  $\Delta d$ ) whether slider phase switches overlap during reversals.

## 7.6. Experimental Validation

### 7.6.1. Force Characterization Experiment

To characterize the thrust force generated by OSCAR, we conducted two complementary sets of experiments using *ex vivo* porcine colon tissue. The first quantified the force cycle of a single slider, allowing us to analyze the contact mechanics at the single slider level. The second measured the net thrust force of the complete capsule

operating inside an intact *ex vivo* colon segment, to connect the slider-level mechanics to the whole-system performance, and validate our theoretical model. The full-robot tests also enabled exploration of how actuation speed and cam design influence the generated thrust.

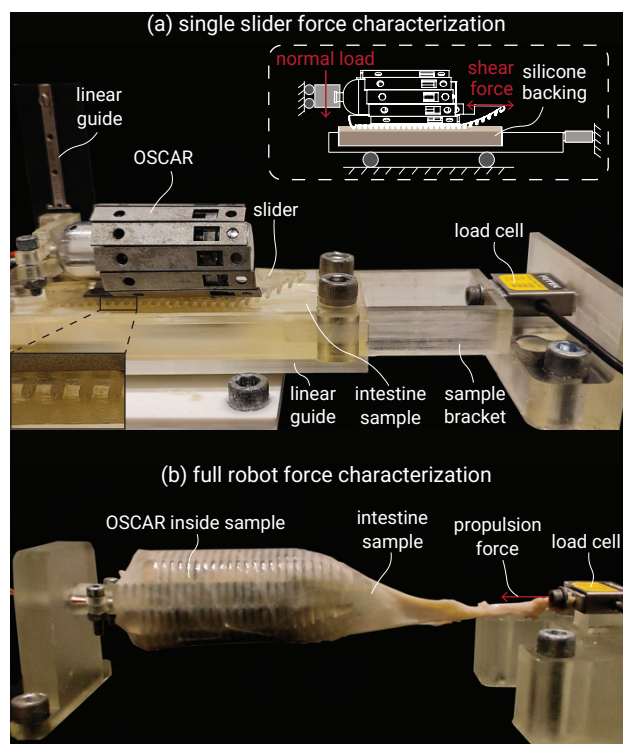
For the single-slider experiment (Fig. 7.7a), we cut open an *ex vivo* porcine colon sample and mounted it flat on a bracket with a silicone-backing layer underneath. We positioned the bracket horizontally on a linear guide and attached it to a co-axial single-axis load cell to measure the tangential forces acting on the tissue. OSCAR was mounted on a vertical guide so that only a single slider was pressed downward onto the tissue under a controlled normal load. Measuring the traction from the tissue side was essential to capture the viscoelastic response of the tissue, which is critical to understand the physics of the slider-tissue interaction. Before each trial, we hydrated the tissue surface with saline and reset the baseline force by lifting and reloading the robot. We repeated each condition six times.

For the full-capsule experiment (Fig. 7.7b), we attached an intact porcine colon segment to a single-axis load cell at one end and inserted the robot coaxially from the other. We fixed the capsule from its base into a mounting bracket to maintain alignment with the load-cell axis. As in the single-slider setup, measuring traction from the tissue side was essential for capturing the wall's viscoelastic response and linking the slider-level mechanics to whole-system behavior. Using this configuration, we evaluated how actuation speed, cam duty cycle, and the number of reversals (i.e., cam jumps) influence the generated thrust force. Before each run, we hydrated the colon specimen externally with saline, and after each trial we relaxed the tissue to release any tension accumulated during the trial. Each experimental condition was performed in six repeated trials.

### 7.6.2. Proof-of-Concept Locomotion Experiment

In this proof-of-concept experiment, we evaluated the capsule's ability to traverse colonic tissue using *ex vivo* porcine colon samples. Testing in real tissue is essential for validating friction-based locomotion systems. Unlike silicone phantoms, which typically exhibit elevated and highly uniform friction, *ex vivo* colon provides a more physiologically representative mechanical properties including viscoelastic and frictional properties which are of great influence on the performance of the capsule as indicated by the model.





**Figure 7.7. Experimental setups for characterizing the thrust force generated by OSCAR.** (a) Single-slider force characterization. An ex vivo porcine colon sample is opened and mounted flat on a silicone-backed bracket, which is guided horizontally and instrumented with a single-axis load cell to measure the traction force from the tissue side. The robot is mounted on a vertical guide so that a single slider presses against the tissue under controlled normal loading. This configuration enables direct observation of the stick-slip traction cycle and captures the viscoelastic response of the wall, which is essential for understanding the slider-tissue contact mechanics. (b) Full-robot thrust characterization. OSCAR is inserted coaxially into an intact porcine colon segment attached to a single-axis load cell. Measuring traction from the tissue side again allows tacking of the traction force on the wall which includes its viscoelastic response, linking individual-slider mechanics to whole-robot propulsion. This setup is further used to evaluate how actuation speed, cam duty cycle, and the number of motion reversals per cycle (cam jumps) influence the net thrust generated by the capsule.

We prepared six ex vivo intestinal tissue samples, each approximately 1 m in length, and immersed them in saline prior to each trial to ensure surface lubricity. The capsule was then introduced at the proximal end of the segment to ensure full wall contact. To impose a more stringent locomotion challenge, the intestinal segments were left in their natural, deflated configuration—without insufflation or suspension—requiring the capsule not only to generate forward thrust but also to pull itself through self-contacted, folded tissue. After activating the capsule, we recorded its progression along the sample until the end of each run.

At the end of every trial, the inner wall of the intestine samples was inspected under a microscope to observe any potential visual tissue damage associated with capsule–wall interaction.

## 7.7. Results and Discussion

### 7.7.1. Single Slider Stick-Slip Contact Mechanics

The recorded tangential force of a single slider against the colonic tissue (Fig. 7.8(a)) revealed their contact mechanics. As the cam drives the slider through the alternating phases of retraction and advance, the measured force switches between positive and negative traction with nearly symmetric peak magnitudes ( $\pm 0.14$  N), which is expected thanks to the symmetric surface design of the slider. The durations of the retraction and advance phases constitute  $\approx 91.1\%$  and  $\approx 8.5\%$  of the time period, respectively. These duration percentages yield a duty asymmetry of  $\Delta d \approx 0.82$ , perfectly matching the duty-cycle asymmetry of the single-jump cam used in this experiment, confirming that the slider is indeed kinematically constrained to the cam profile.

Examining the measured force profile in Fig. 7.8(a) reveals distinct behaviors in the two motion phases. During retraction, the traction on the wall decreases (increases from the capsule's perspective) linearly up to a threshold of  $\approx -0.08$  N, beyond which the force slowly converges to the retraction friction limit  $-0.14$  N following an exponential decay. In contrast, the advance phase exhibits an approximately linear increase (decrease from the capsule's perspective) in traction until the advance friction limit  $0.14$  N (negative thrust from the capsule's perspective) is reached. These smooth reversals between peak positive and negative traction reflect the viscoelastic response of the colonic wall, whose deformation and time-dependent relaxation shape the contact mechanics. Since the force is measured from the tissue wall side (Fig. 7.7(a)), this viscoelastic behavior is fully captured in the experimental data.

The observed contact behavior (linear followed by exponential convergence in retraction and linear in advance) can be interpreted through the mechanics of stick and slip. During stick, the slider and the colonic wall move together, with no relative velocity ( $v_{\text{rel}} = 0$ ), and therefore the tissue is kinematically constrained to follow the slider displacement. As the slider moves, the wall stretches tangentially, building up an internal viscoelastic force governed by the elastic and viscous properties of the tissue.

In this displacement-driven regime, the tangential traction experienced by the slider is equal and opposite to the elastic restoring force of the tissue, producing an approximately linear increase or decrease in traction depending on the slider motion. Stick persists until this internal tissue force reaches the static friction limit ( $F_{tissue} \geq \mu_s N$ ).

Once the accumulated internal force exceeds the static friction threshold, the slider force can no longer stretch the tissue wall further, and thus, the slider starts to slip relative to the wall ( $|v_{rel}| > 0$ ). In this regime, the slider thrust force saturates at the kinetic friction limit  $\mu_k N$ , while the tissue is no longer kinematically constrained to the slider. Instead, its internal viscoelastic state evolves under a force-driven condition. The tissue relaxes from its previously stored deformation toward the equilibrium set by the driving kinetic friction force. This relaxation follows an exponential decay with a characteristic time constant governed by the viscoelastic properties of the tissue, producing the gradual convergence toward the friction limit observed in the measurements. As a result, the wall displacement and internal force evolve continuously during slip, while the slider thrust force remains fixed at the kinetic friction limit. Slip continues until the internal tissue force becomes smaller than the kinetic friction limit ( $F_{tissue} < \mu_k N$ ), after which contact returns back to the stick regime. This stick–slip dynamics can be mathematically described as follows:

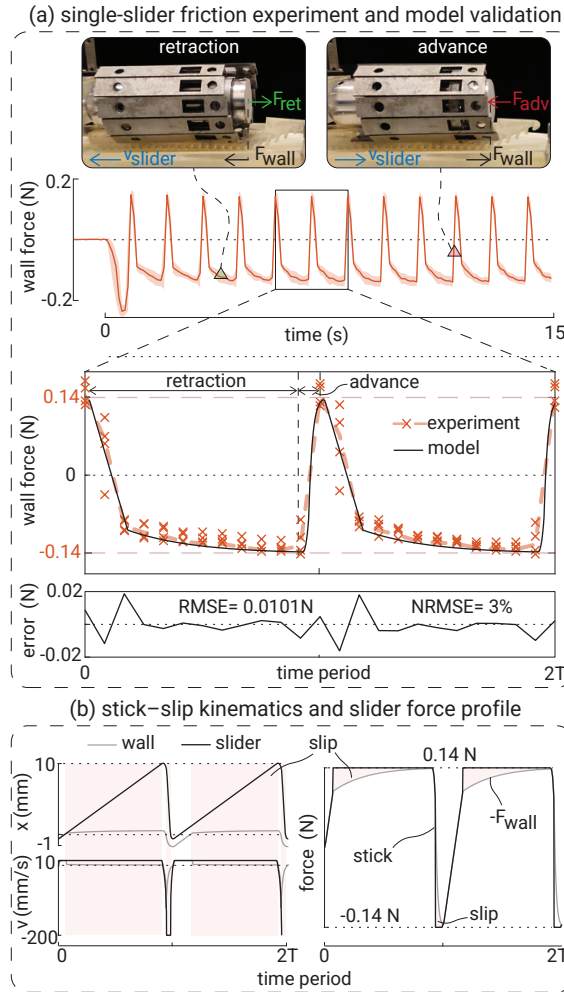
$$F_{slider} = \begin{cases} -F_{tissue}, & |F_{tissue}| < \mu_s N, (\text{stick}) \\ \mu_k N \operatorname{sgn}(v_{rel}), & |F_{tissue}| \geq \mu_k N, (\text{slip}) \end{cases} \quad (7.19)$$

where  $F_{tissue}$  is the internal force of the tissue,  $N$  is the normal load,  $\mu_s$  and  $\mu_k$  are the static and kinetic coefficients of friction, respectively, and  $\operatorname{sgn}(v_{slider})$  is the sign of the slider velocity.

To capture the viscoelastic response of the colonic wall during both stick and slip, we model the tangential tissue mechanics as a single in-parallel spring–dashpot element, also known as Kelvin–Voigt model [62, 65, 66], characterized by an effective elongation stiffness  $k_{tissue}$  and damping coefficient  $c_{tissue}$ . During stick, the wall is kinematically constrained to follow the slider, and thus the internal force of the tissue can be obtained directly from the viscoelastic resistance to motion, expressed as follows:

$$F_{tissue}(t) = k_{tissue} x_{wall}(t) + c_{tissue} \dot{x}_w(t), (\text{stick}) \quad (7.20)$$

where  $x_{wall}(t)$  is the displacement of the wall. Once slip initiates and the interface becomes force-driven, the tissue is no longer constrained by the slider kinematics.



**Figure 7.8. Single-slider friction experiment and validation of the viscoelastic stick-slip model.**

(a) Experimental setup and measured wall-force response during alternating retraction and advance of a single slider. The recorded traction force exhibits symmetric peak traction ( $\pm 0.14$  N) and clear stick-slip transitions shaped by the wall's viscoelastic response. The zoomed-in view of the traction force profiles show the linear build-up during stick and the exponential convergence toward the friction limit during slip, together with the model prediction obtained using the experimentally-identified elongation stiffness  $k_{tissue} \approx 130 \text{ N.m}^{-1}$  and viscous damping  $c_{tissue} \approx 6 \text{ N.s.m}^{-1}$  parameters. The bottom panel shows error profile over two periods, with the model achieving  $\text{RMSE} = 0.0101 \text{ N}$  and  $\text{NRMSE} \approx 3.6\%$ . (b) Estimated wall and slider kinematics and the corresponding stick-slip force evolution. The model reconstruction shows how slider and wall motion coincide during stick, producing linear traction growth, and how the slider force saturates during slip while the wall relaxes according to the tissue's viscoelastic dynamics. These results demonstrate that the stick-slip formulation accurately reproduce the recorded interaction and reveal the underlying contact mechanics. Refer to Supplementary video.2 for a video demonstration of the single slider force characterization experiment.

**Algorithm 1** Stick-slip capsule-wall interaction model**Require:** sampled time series  $t(n)$ ,  $\Delta t$ ,  $x_{slider}$ ,  $k_{tissue}$ ,  $c_{tissue}$ ,  $\mu_s$ ,  $\mu_k$ ,  $N$ ,  $\tau_{ret}$ ,  $\tau_{adv}$ 

```

1:  $mode \leftarrow \text{STICK}$ 
2: for  $i = 2$  to  $n$  do
3:    $\Delta x_{slider} \leftarrow x_{slider}(i) - x_{slider}(i-1)$ ;  $v_s \leftarrow \Delta x_{slider} / \Delta t$ 
4:    $\tau \leftarrow \tau_{ret}$  if  $v_s \geq 0$  else  $\tau_{adv}$ ;  $\alpha \leftarrow 1 - e^{-\Delta t / \tau}$ 
5:   if  $mode = \text{STICK}$  then
6:      $x_{wall}(i) \leftarrow x_{wall}(i-1) + \Delta x_{slider}$ 
7:      $v_{wall}(i) \leftarrow (x_{wall}(i) - x_{wall}(i-1)) / \Delta t$ 
8:      $F_{wall}(i) \leftarrow k_{tissue} x_{wall}(i) + c_{tissue} v_{wall}(i)$ 
9:     if  $|F_{wall}(i)| \geq \mu_s N$  then;  $mode \leftarrow \text{SLIP}$ 
10:    end if
11:  else if  $mode = \text{SLIP}$ 
12:     $F_{target} = \mu_k N \text{sign}(v_{wall}(i))$ 
13:     $F_{wall}(i) \leftarrow F_{wall}(i-1) + \alpha (F_{target} - F_{wall}(i-1))$ 
14:     $x_{target} = F_{wall}(i) / k_{tissue}$ 
15:     $x_{wall}(i) \leftarrow x_{wall}(i-1) + \alpha (x_{target} - x_{wall}(i-1))$ 
16:     $v_{wall}(i) \leftarrow (x_{wall}(i) - x_{wall}(i-1)) / \Delta t$ 
17:    if  $|F_{wall}(i)| < |k_{tissue} x_{wall}(i)|$  then;  $mode \leftarrow \text{STICK}$ 
18:    end if
19:  end if;  $F_{elastic}(i) \leftarrow x_{wall}(i) k_{tissue}$ 
20: end for; return  $x_{wall}$ ,  $v_{wall}$ ,  $F_{wall}$ ,  $F_{elastic}$ 

```

Therefore, its internal force, and consequently its deformation, relaxes toward the friction-limited driving force  $\mu_k N \text{sgn}(v_{slider})$ . This relaxation in force and displacement can be captured with the Kelvin–Voigt first-order relaxation law as follows:

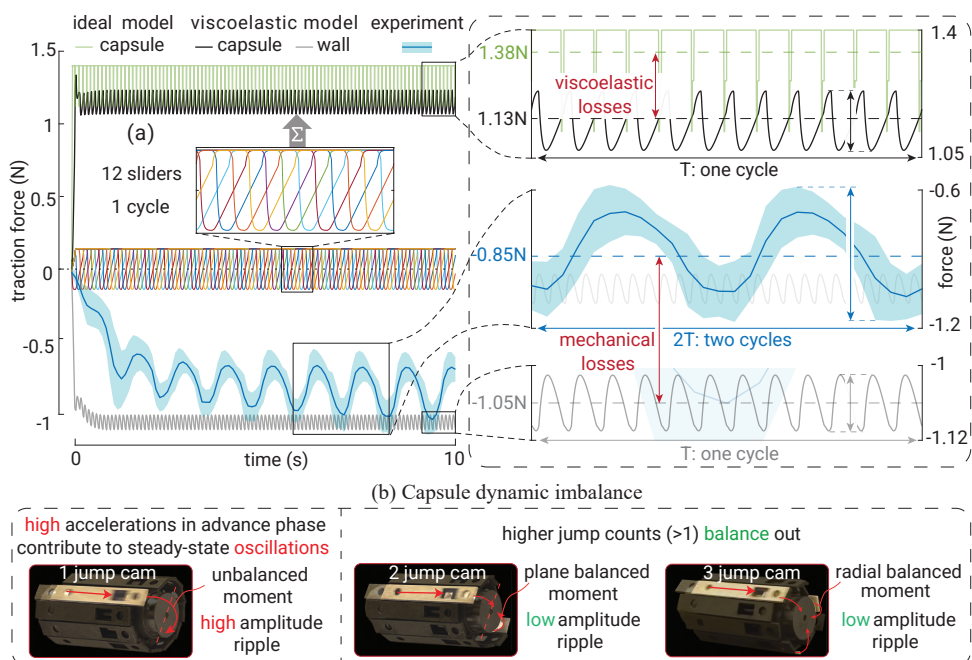
$$\begin{aligned} \dot{F}_{tissue}(t) &= -\frac{1}{\tau(V)} (F_{tissue}(t) - F_{slider}(t)), \\ \dot{x}_w(t) &= -\frac{1}{\tau(V)} \left( x_w(t) - \frac{F_{tissue}(t)}{k_{tissue}} \right), \text{ (slip)} \end{aligned} \quad (7.21)$$

with  $\tau$  representing the effective relaxation time constant of the tissue. The elongation stiffness  $k_{tissue}$  and viscous coefficient  $c_{tissue}$  of the tissue wall were identified directly from the single-slider force recordings (Fig.7.8(a)) by fitting the stick portions of the cycle. In this displacement-driven regime, the measured traction follows the Kelvin–Voigt resistance to the imposed wall motion, and thus the linear traction build-up during stick provided  $k_{tissue}$  and  $c_{tissue}$  in Eq. (7.20), which we identified to be  $\approx 130 \text{ N m}^{-1}$  and  $\approx 6 \text{ N s m}^{-1}$ , respectively, in both phases of advance and retraction.

The exponential convergence toward the kinetic-friction limit during slip then provided the effective relaxation time constant  $\tau(v)$  in Eq. (7.21). The convergence showed a rate-dependent relaxation time constant  $\tau(v)$ , yielding  $\tau \approx 0.02$  s during advance and  $\tau \approx 0.45$  s during retraction, consistent with previously reported rapid relaxation rates of intestinal tissue at higher strains [62, 67]. Additionally, the faster relaxation time  $\tau$  at higher actuation speed (advance) is consistent with previously reported rate-dependent viscoelastic relaxation in intestinal tissue [64, 68, 69].

Using the experimentally identified viscoelastic parameters together with the established stick–slip formulation detailed in Algorithm 1, we were able to reproduce, with high fidelity, the measured stick-slip behavior of the single-slider interaction with the tissue wall. The model captures the key qualitative features of the force response, including the linear force build-up during stick, the smooth transitions into slip, and the characteristic viscoelastic relaxation observed experimentally. Quantitatively, the model achieved a root-mean-square error of  $\text{RMSE} = 0.0101$  N across cycles, corresponding to a normalized error of  $\text{NRMSE} \approx 3.62\%$  of the full force range, as shown in Fig. 7.8(a), which is below the typical range reported for viscoelastic soft-tissue models, where residuals of 5 – 15% are common due to inherent biological variability [65]. This close agreement confirms that the dominant physics of the slider–tissue interaction are well explained by our viscoelastic stick-slip formulation, validating its utility for the multi-slider thrust model.

With the stick–slip formulation and viscoelastic parameters validated against the measured force profile, we can now reconstruct the full kinematics of the wall and the corresponding forces acting on both the slider and the tissue. As shown in Fig. 7.8(b), the model shows how the wall displacement follows the slider displacement during stick, producing the linear traction build-up, and how it subsequently relaxes during slip as the slider force saturates at the friction limit. During stick, the slider and wall forces follow one another exactly, reflecting their kinematic constraint, whereas during slip the slider force remains fixed at the kinetic friction limit while the tissue wall force relaxes according to the tissue’s viscoelastic relaxation. This mechanics reconstruction provides a clear picture of the local stick-slip interaction at the level of a single slider, providing the basis to understand the collective behavior of the capsule.



**Figure 7.9. Capsule thrust force and dynamic imbalance.** (a) Comparison between the measured traction force generated by the capsule on the ex vivo colon wall (blue), the wall and capsule thrust force estimated by the viscoelastic model (gray and black), and the capsule thrust force estimated by the ideal model (the friction-limited case) (light green). The ideal model predicted cycle-averaged capsule thrust force (1.38 N) compared to (1.13 N) predicted by the viscoelastic model, which reflects the losses due to wall viscoelasticity. The absolute value of the experimentally measured steady-state average traction ( $\approx 0.85$  N) is less than the absolute value of the wall force predicted by the viscoelastic model (1.05 N), attributed to the mechanical losses in the system. Insets show model–experiment comparison over one and two cam cycles, highlighting viscoelastic losses (top) and system-level mechanical losses (bottom). (b) Illustration of the inertial moment imbalance generated by sliders during the rapid advance phase. In the single-jump cam, only one slider undergoes the high acceleration motion reversal when switching to advance phase without any balancing counter force, producing a net rocking moment on the capsule and a large-amplitude ripple in the measured thrust. For higher jump counts, sliders advance in symmetric groups (two opposite sliders in the two-jump cam, three evenly spaced sliders in the three-jump cam), balancing the inertial moments and substantially reducing the thrust ripple. Refer to supplementary video.3 for a video demonstration of the full capsule force characterization experiment.

### 7.7.2. Capsule Thrust Mechanics

Building on the validation of the single-slider contact mechanics, we evaluated the traction of the full capsule using the same (single jump) cam profile, allowing for a direct comparison between the measured thrust and the predictions of the validated stick-slip model. The full capsule force experiment showed a mean steady-state traction force on the tissue wall of  $\approx -0.85$  N (see Fig. 7.9(a)), confirming the capsule's ability to generate steady thrust. The measured steady-state wall traction deviates

from the  $-1.05$  N predicted by the viscoelastic model, attributed to the mechanical losses associated with the full system assembly, which are not accounted for in the model. Additionally, the corresponding capsule thrust force computed using our viscoelastic model was  $1.13$  N, compared to  $1.38$  N predicted by the model in the ideal case of the contact force always saturating at the kinetic friction limit. The deviation in this case arises from the viscoelastic properties which alter the force profile waveform, yielding a different net cycle-averaged thrust.

The recorded traction force on the wall also showed steady-state oscillations (ripple) with an amplitude of  $\approx 0.3$  N, as shown in Fig. 7.9(a). While our model predicts that the capsule traction force, and consequently the force on the tissue wall, will exhibit oscillations stemming from the instantaneous overlapping of sliders switching from retract to advance motion, the experimental data diverge from these predictions. The observed amplitude was six times greater than the ripple amplitude predicted by the viscoelastic model. Furthermore, the experimental frequency matched the actuation frequency (one peak per cycle), whereas the model predicted a frequency of 12 peaks per cycle, corresponding to one motion reversal of each of the twelve sliders per one cycle. This mismatch in both amplitude and frequency suggests that the source of the observed ripple in the experiment differs from the mechanism assumed in the model.

The unexpectedly large ripple in the full-capsule force can be explained by a small rocking moment during the steep advance phase. Although the slider is kinematically constrained to the cam, its rapid acceleration produces an inertial force with an axial component, which generates a tilting moment on the capsule (Fig. 7.9(b)) and induces a small rotation  $I\ddot{\phi} = M$ . Even sub-degree tilts can create axial displacements that, with the measured wall stiffness  $k_{\text{tissue}}$ , lead to force oscillations consistent with the  $\sim 0.6$  N peak-to-peak ripple. The effect is strongest for the single-jump cam due to the unbalanced rocking moment, and is reduced for higher jump-count cams where multiple sliders simultaneously advance, which balance the capsule and partially cancel out the rocking moment (Fig. 7.9(b)).

### 7.7.3. Influence of Actuation Speed and Duty Asymmetry

Examining the role of the actuation speed on thrust revealed that the generated thrust remained effectively unchanged across the input voltages tested. As shown in Fig. 7.10(a), operating the motor at 6 V, 9 V, or 12 V, corresponding to three different actuation speeds, produced nearly identical steady-state averaged traction



forces of  $\approx 0.85$  N, with the time–evolution profiles closely overlapping, indicating that the capsules’ thrust is speed-independent. This speed independence confirms the validity of the Coulomb friction law, used in the theoretical model, within the operating speeds tested. Furthermore, it shows that the generated thrust force is robust against speed variations that can result from varying normal loading.

Modifying the cam geometry produced systematic and predictable changes in the mean thrust. Increasing the number of motion reversals (jumps) from one to two and three jumps decreased the cycle-averaged traction force, as shown in Fig. 7.10(a). This decrease follows directly from the reduced duty asymmetry  $\Delta d$  from 0.82 to 0.64 and 0.48 of the single, double, and triple jump(s) cams, respectively. As the duty asymmetry decreases, a larger fraction of the sliders spend more time in the advance phase than in retraction, thereby reducing the generated net thrust. This trend is consistent with the analytical predictions of Eq. 7.14 as depicted in Fig. 7.6(a), in which the model anticipates a monotonic dependence of the averaged thrust force on the duty-cycle asymmetry. As shown in Fig. 7.10(b), the model closely matches the recorded traction force once both viscoelastic effects and system-level mechanical losses are incorporated.

## 7

However, while the model matches the trend of the recorded average traction force, the observed ripple in the recorded traction waveform does not follow the model’s prediction. The model predicts the largest ripple amplitude for the three-jump cam and the smallest for the single-jump cam, due to the increased number of slider phase switching as the jump count increases. Experimentally, the opposite occurred: as shown in Fig. 7.10(b), the single-jump cam exhibited the highest ripple amplitude, whereas the two- and three-jump cams showed progressively less ripple amplitude in their recorded traction force profiles. This discrepancy can be explained by the dynamic imbalance of the full capsule, depicted in Fig. 7.9(b). The effect is expected to be most pronounced for the single-jump cam, and reduced for the two- and three-jump cams as the advancing-slider motion becomes more evenly distributed and balanced around the circumference of the capsule.

The close qualitative and quantitative agreement between the model and the experiments in traction-force behavior across both speed variations and cam geometries supports the theoretical conclusion that OSCAR’s self-propulsion is governed primarily by duty-cycle asymmetry and the viscoelastic stick–slip behavior at each slider–wall interface, independent of speed. Additionally, the ripple observed in the experiments underscores the importance of phase distribution and synchronization

of sliders, and the dynamic balance of the capsule's motion cycle in achieving smooth traction force.

#### 7.7.4. Proof-of-Concept Locomotion Demonstration

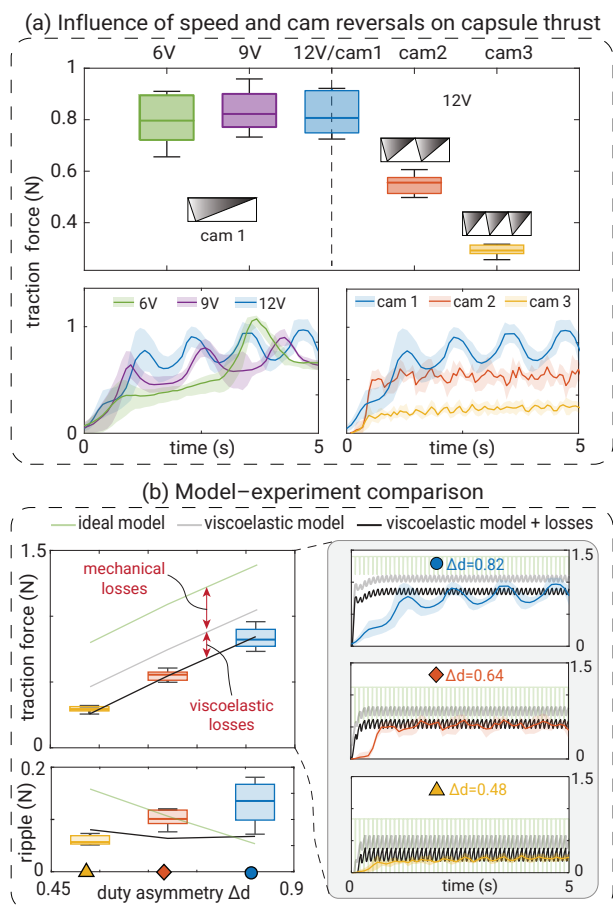
To demonstrate the self-propelling capability of OSCAR, we conducted locomotion validation experiments using *ex vivo* porcine colon samples, in which we inserted the capsule into intact samples of the *ex vivo* porcine colon and activated the capsule, as shown in Fig. 7.11(a). Upon activation, the capsule successfully generated sufficient thrust to advance forward, effectively reopening the naturally collapsed lumen as it progressed (Fig. 7.11(b)). Across six samples, OSCAR successfully traversed the full 100 cm length at an average speed of  $3.08 \pm 0.08$  mm/s. Extrapolating this velocity to the average human colon length of 1.6 m suggests a potential total intubation time of approximately eight minutes, a duration comparable to that of conventional colonoscopy [36]. Furthermore, post-trial microscopic inspection of the mucosal surface showed no visible damage to the tissue wall (Fig. 7.11(b)). This demonstration highlights the potential of OSCAR to achieve clinically relevant locomotion speeds while maintaining safe interaction with the colon wall.

#### 7.7.5. Implications on Future Designs and Generality of Results

While this study focused on introducing OSCAR and validating its working principle, the theoretical insights gained here provide general principles that go beyond the specific design of OSCAR.

On a system level, the duty-cycle framework used in this work to describe the retraction and advance phases of the cam profile is a robust, device-agnostic tool. It shows that locomotion fundamentally depends on the asymmetry between the “grip” and “release” phases, regardless of the actuation source. This abstraction therefore provides a transferable metric for future designs, whether driven by cams, linear actuators, or soft robotic transmissions.

On an interface level, the stick-slip model presented in Algorithm 1 sheds light on what constitutes an ideal capsule design for systems that rely on sliding friction to generate locomotion. In this context, the ideal sliding-friction capsule is one in which sliders slip for most of the cycle, making thrust less sensitive to wall viscoelasticity. According to the stick-slip model, this can be achieved by increasing the stroke height of the slider (resulting in higher elastic resistance  $k \cdot x_{slider}$  of the wall) and/or



**Figure 7.10. Effect of speed and cam jumps on capsule thrust, and comparison between models and experiments.** (a) Cycle-averaged traction forces generated by OSCAR under different voltages (6V, 9V, 12V) corresponding to different motor speeds using the single-jump cam, and under the three different cam geometries with (1, 2, 3) cam jumps at 12V driving voltage. Box plots and the corresponding time-evolution traces show the mean thrust remained approximately constant across the different speed, whereas increasing the number of cam jumps reduces the net traction and produces progressively smoother force profiles due to improved moment balancing across sliders. (b) Comparison between measured traction, the ideal friction model, the viscoelastic model, and the viscoelastic model with subtracted mechanical losses, plotted against duty-cycle asymmetry  $\Delta d$ . Viscoelasticity lowers the ideal thrust, and additional mechanical losses observed in the full system shifts the prediction further downward, closely matching the experimental data. The lower-left panel shows corresponding ripple amplitudes. Right: the corresponding time-evolution traces for each cam design demonstrate how model and experiment agree in mean thrust while differing in ripple due to system-level inertial effects.

by increasing slider velocity  $v_{\text{slider}}$  through steeper motion profiles (resulting in higher viscous resistance  $c \cdot v_{\text{slider}}$  of the wall), both of which help the sliders break the stick phase and overcome the viscoelastic force of the wall, enabling quick transition to the

slip mode. For a sliding-friction capsule to self-propel, it must be able to transition from stick to slip during its motion cycle. If the capsule remains in stick mode, the colon wall will stretch back and forth with the sliders, trapping the capsule and thus preventing it from propelling forward.

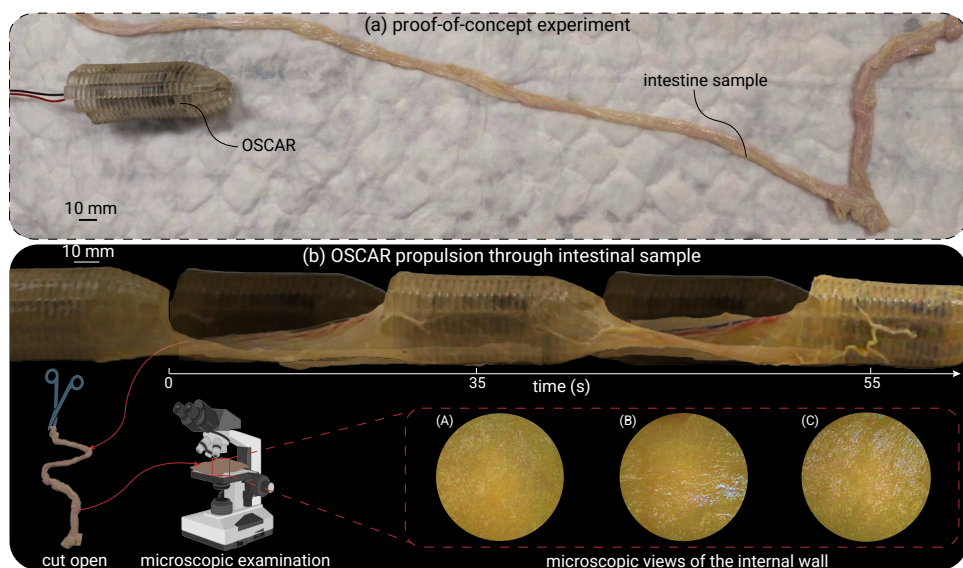
Finally, the stick–slip model presented in this study provides a general tool to simulate capsule designs that rely on sliding friction to generate thrust. The model takes the slider system kinematics and tissue viscoelastic properties as inputs, and predicts the capsule–wall interaction, the resulting capsule thrust, and the associated wall deformation and internal forces. Since we validated the model experimentally, it can be used to guide future iterations of OSCAR and other sliding-friction capsule designs by enabling simulation-based design optimization and performance prediction prior to implementation.

## 7.8. Limitations and Future Directions

While the presented results demonstrate OSCAR's ability to self-propel and its underlying working principle, several limitations and challenges remain to be addressed to optimize the design and performance of OSCAR, and translate it into a fully operational clinical platform.

The current capsule prototype was designed to validate the locomotion working principle and study the factors that affect its performance, resulting in a self-contained architecture with limited available internal volume, which limited the ability to integrate essential clinical payloads such as a camera and biopsy tools. Future iterations will address this constraint by potentially relocating the motor outside the capsule and transmitting motion through a flexible drive shaft, thus freeing internal volume for tool integration. Additional space could also be gained by reducing the number of sliders. Although this would reduce the thrust capacity of the capsule, the loss could be compensated for by integrating active friction-control methods, such as ultrasonic friction modulation [70] or electroadhesion [71], to control the friction of individual sliders with the wall, which could potentially increase the effective friction anisotropy, and thus increase the average thrust.

The current cam mechanism generates unidirectional forward propulsion. While sufficient for cecal intubation, clinical workflows require bidirectional propulsion for better controllability and fine inspection during withdrawal. Future designs will incorporate a switchable or dual-profile cam system to enable bi-directional locomotion.



**Figure 7.11. ex vivo validation of locomotion and tissue safety in porcine intestine.** (a) Proof-of-concept experimental setup showing OSCAR alongside a long, excised porcine colon sample in its natural, collapsed configuration (without insufflation). (b) Time-lapse visualizing the progression of the capsule through the colon sample over a 55-second interval. The robot successfully propelled forward through the colon samples, achieving an average speed of 3.08 mm/s. This speed corresponds to a theoretical full-length colon traversal time of  $\approx 8$  minutes, comparable to the standard for conventional colonoscopy. Following locomotion, the intestinal tract was dissected (schematic inset) to assess the integrity of the tissue wall. (A–C) Representative microscopic views of the internal mucosal wall at distinct locations along the traversal path. The images reveal an intact epithelial surface with no visible abrasions, hematomas, or disruption, confirming the atraumatic nature of the ovipositor-inspired self-propelling mechanism. Refer to supplementary video.4 for video demonstration of the locomotion validation experiment.

Additionally, geometric adaptability is required to handle the high variability of diameters and shapes along the different segments of the colon. The integration of radially compliant or expandable sliders, similar to those proposed in [40, 56], would allow the capsule to maintain optimal wall contact pressure across varying lumen sizes and shapes.

Another limitation of the current study is that all experiments were performed under ambient conditions in straight *ex vivo* colon segments. *In-vivo*, the colon is highly tortuous, variable in size and shape, with varying boundary conditions along its length and subject to intra-abdominal pressure, all of which can influence the capsule's thrust capacity. For example, the intra-abdominal pressure acting on the colon wall can potentially increase the normal contact force between the capsule and the wall, which can both increase the frictional grip and the viscoelastic damping of the wall. Additionally, curved segments can introduce additional drag, particularly through the

tether of the capsule which becomes increasingly significant as the capsule goes deeper into the colon. Future work will therefore extend both the experiments and the analytical model to evaluate OSCAR's thrust capacity and drag in curved, pressurized, and geometrically-constrained conditions.

Finally, from a sterility perspective, the current slider system includes small gaps between neighboring sliders, which may trap biological residue, posing clinical risks and sterilization challenges. To satisfy clinical sterilization standards, future designs will explore encapsulating the mechanism within a continuous, deformable medical-grade skin, eliminating the existing gaps between sliders and potentially improving sterility. Such a continuous skin could potentially also enhance the self-propelling performance of the capsule by increasing the effective movable contact area with the mucosa.

## 7.9. Conclusion

We introduced OSCAR, an ovipositor-inspired self-propelling capsule robot designed to overcome the fundamental traction limitations of soft-tissue locomotion and reduce the discomfort of conventional colonoscopy by mimicking how parasitic wasps coordinate their ovipositor valves to transport eggs into a host substrate. OSCAR mechanically encodes the ovipositor-inspired motion pattern through a spring-loaded cam system that drives twelve circumferential sliders in a phase-shifted reciprocating pattern, generating a controlled friction anisotropy at the interface that enables OSCAR to self-propel through the slippery, viscoelastic environment of the colon without requiring excessive normal loads. Experimental validation in *ex vivo* porcine colon demonstrated a steady-state traction force of  $\approx 0.85$  N and an average speed of 3.08 mm/s, highlighting the potential of OSCAR to match standard safety and cecal intubation times of conventional colonoscopy.

Additionally, we presented an analytical framework that links cam kinematics to capsule thrust by capturing the viscoelastic stick-slip interaction at the tissue interface using a Kelvin-Voigt formulation, and validated this framework experimentally. The model revealed that thrust is largely speed-independent and scales linearly with cam duty-cycle asymmetry, underscoring OSCAR's predictable performance and scalable design. Beyond OSCAR, the model provides a robust, device-agnostic tool to simulate and optimize future sliding-friction capsule designs.

By tuning friction anisotropy through mechanically-encoded motion patterns of sliders, OSCAR establishes a pathway toward friction-programmable robotic capsules that generate robust and predictable locomotion in the challenging environment of the colon and beyond, ultimately bringing painless robotic colonoscopy within reach.

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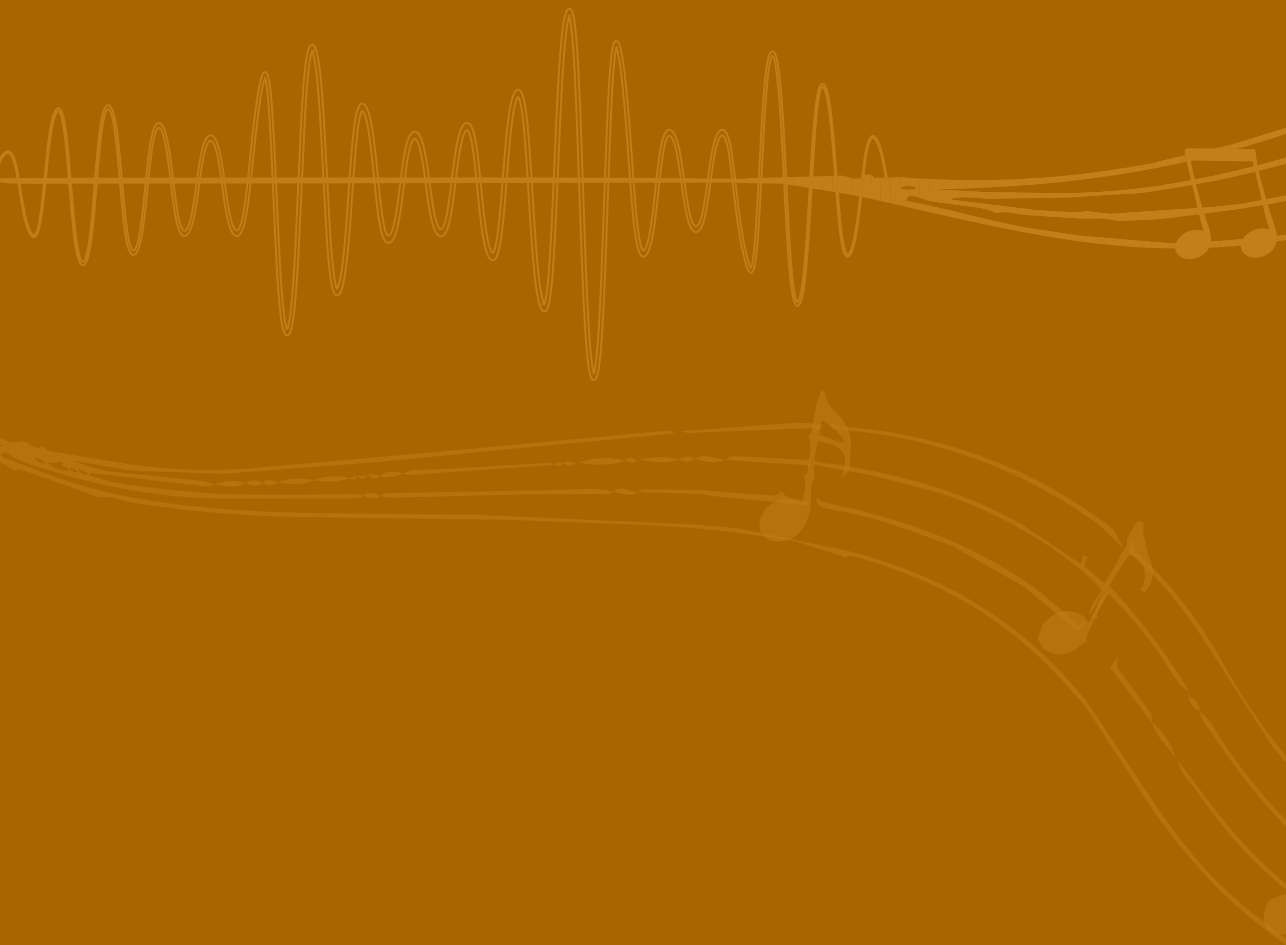
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# PART III

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# Glide and Grip

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## to Propel





# CHAPTER 8

## Ultrasonic Lubrication for Robotic Locomotion

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## Abstract

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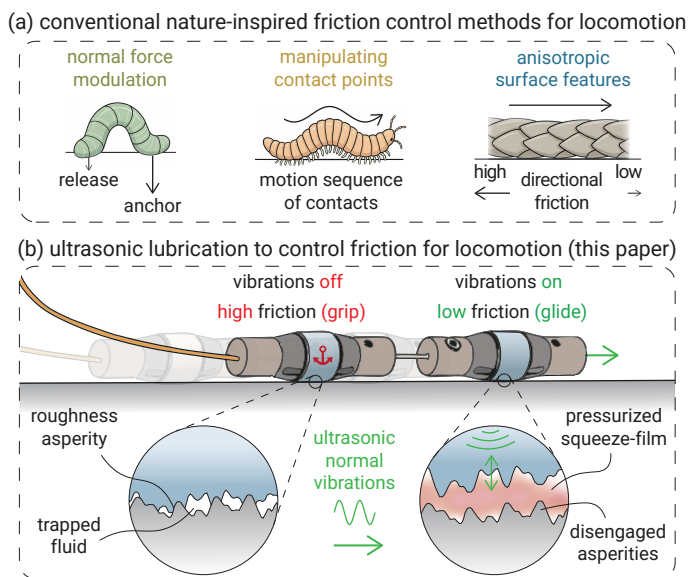
Friction is the essential mediator of terrestrial locomotion; yet, in robotic systems, it is usually treated as a passive property fixed by surface materials and conditions. This chapter combines the local-level friction control via ultrasonic lubrication, developed in Part I, with the system-level friction anisotropy tuning via motion coordination of Part II, to enable active friction control in multi-contact robotic locomotion systems. Unlike conventional approaches that rely on contact loading, passive surface anisotropy, or motion coordination to generate frictional asymmetry, this approach uses ultrasonic lubrication to control friction independently at individual contact interfaces to generate the friction differential required for locomotion. To validate this concept, we developed two friction-control modules based on ultrasonic lubrication: a cylindrical design for lumen-like environments and a flat-plate design for external surfaces, and integrated them into two bio-inspired multi-contact locomotion systems modeled after inchworm and wasp ovipositor gaits. In both systems, ultrasonic lubrication enabled bidirectional locomotion using only two contact elements by selectively reducing friction at the friction-control modules, achieving locomotion efficiencies exceeding 90%, whereas the same systems failed to achieve locomotion without ultrasonic lubrication. Friction characterization further showed substantial friction reduction across rigid, soft, granular, and biological tissue interfaces under dry and wet conditions and across different roughness levels, highlighting the robustness and versatility of ultrasonic lubrication as a route toward simpler and more efficient robotic locomotion systems.

## 8.1. Introduction

Biological organisms achieve remarkable locomotion by exploiting frictional interactions at the body–environment interface. Inchworms advance through sequential frictional anchoring of body segments combined with extension–contraction cycles, while earthworms propel through retrograde peristaltic waves aided by ventral setae to modulate frictional grip along the body [2–5]. Parasitic wasps steer ultra-thin ovipositors through dense substrates using reciprocating interlocking valves (sliders) that create frictional anisotropy, enabling substrate penetration without buckling [6–8], and limbless organisms, such as snakes, utilize overlapping ventral scales to create directional frictional asymmetry that enables them to propel. Fundamentally, these biological systems control friction at the contact interface to achieve locomotion through three primary strategies: by modulating the normal force at the contact points to anchor or release, by coordinating the motion sequence of those contact points to generate friction differential, or by employing anisotropic surface features to create directional friction asymmetry, as illustrated in Fig.8.1(a).

These locomotion strategies have inspired a wide range of robots for confined and challenging environments, from peristaltic soft crawlers for search-and-rescue in rubble [3, 9, 10], to earthworm-inspired robots for pipeline inspection [11–13], to inchworm-inspired robots for medical applications such as self-propelling endoscopic capsules [14–16]. Across these developments, robotic systems have largely sought to replicate the friction-control principles observed in biological systems. To emulate normal-force modulation, inflatable fluidic structures have been used to vary contact pressure and thereby control the anchoring and release of robot segments [14, 17]. To reproduce contact sequencing, cam-driven mechanisms and fluidic control systems have been employed to coordinate the motion of multiple sliders and generate the desired friction differential [7, 18, 19]. In parallel, directional pads with scale-like features have been introduced to create directional frictional asymmetry [4, 20–23].

However, these friction-control techniques impose inherent limitations in the design of locomotion systems. Strategies based on modulating normal force require the robot to locally expand or apply elevated contact forces to anchor, which can limit adaptability in fragile or tightly confined environments and often require dedicated actuators. Strategies based on coordinating the motion sequence of multiple contacts demand precise gait control through dedicated actuators and/or sophisticated



**Figure 8.1. Concept of ultrasonic lubrication for locomotion.** (a) Conventional nature-inspired friction control methods for locomotion through normal force modulation, contact sequencing, or anisotropic surface features. (b) The proposed approach uses transverse ultrasonic vibrations to actively control friction at the contact interface. These vibrations pressurize the trapped fluid between the microscopic roughness asperity junctions, creating a pressurized fluid cushion that disengages the two surfaces, allowing for controllable switching between high-friction ‘grip’ (vibrations off) and low-friction ‘glide’ (vibrations on) states to drive locomotion.

motion-transmission systems and often require a large number of contacts to generate sufficient friction differential for traction, significantly increasing the overall system complexity. Strategies based on anisotropic surface features rely either on fixed microscopic textures that cannot adapt to changing locomotion demands, or on controllable macroscopic scales that further complicate the design by requiring onboard mechanisms for their deployment and actuation. These limitations suggest the need for an alternative friction-control method that can regulate the frictional properties of the interface independently of normal load, number of contacts and surface texture.

We present ultrasonic lubrication as an active friction control method for robotic locomotion. Unlike approaches that vary normal force or use surface anisotropy, ultrasonic lubrication directly modulates the coefficient of friction, independently of normal load, contact count, and surface design. This method uses transverse ultrasonic vibrations at the interface to generate a pressurized fluid film that lubricates the contact interface, thereby reducing the coefficient of friction for smooth low-friction sliding on demand, as illustrated in Fig. 8.1(b). By dynamically adjusting the coeffi-

cient of friction at the body-environment interface, a locomotor can determine when and where to grip or glide, moving beyond the limits of passive interfaces and opening up new possibilities for more efficient and less mechanically complex robotic locomotion systems. To investigate the viability of this approach, we present two ultrasonic lubrication module designs, which we implement and integrate into two bio-inspired locomotion systems inspired by inchworm and ovipositor gaits, respectively, and evaluate their performance under different locomotion conditions.

In the remainder of this chapter, Section 8.2 introduces the working principle of ultrasonic lubrication. Section 8.3 presents the design, fabrication, and integration of two friction control modules. Section 8.4 describes the experimental setup and protocols, and Section 8.5 reports the vibration, locomotion, and friction modulation results. Finally, Section 8.6 summarizes the findings and discusses their implications for future robotic locomotion systems.

## 8.2. Working Principle

### 8.2.1. Ultrasonic Lubrication

When two solid bodies come into contact and slide against each other, friction emerges due to the interaction of a discrete number of microscopic surface asperities, which form micro-junctions at the interface. The sum of the surface areas of these micro-junctions constitutes the real contact area, which is much smaller than the apparent area of contact. As a result, a large fraction of the apparent contact area does not engage directly, leaving fluid from the surrounding environment trapped at the interface (Fig.8.1(b)) that form a thin interfacial layer, commonly described as a squeeze film.

Introducing transverse ultrasonic vibrations to the contact bodies causes the trapped fluid to undergo nonlinear compression, generating excess pressure within the fluid film. This positive pressure separates the two surfaces, disengaging the roughness asperities and lubricating the interface, thus reducing the coefficient of friction, a mechanism known as ultrasonic lubrication or ultrasonic friction modulation [24, 25]. The degree of separation between the two surfaces and therefore the lubrication effect is proportional to the vibration amplitude, allowing precise and active control of the coefficient of friction at the interface on demand [24].

Ultrasonic lubrication has been successfully applied in a range of contexts where friction control is critical, including tactile displays [26], non-contact object manipulation [27], and squeeze-film bearings [28, 29]. Furthermore, it has been shown to operate effectively in submerged liquid environments [30, 31], highlighting its versatility across diverse operating conditions. This ability to switch between high- and low-friction states in different environments and conditions on demand suggests the potential of ultrasonic lubrication for applications requiring active friction control, such as robotic locomotion.

### 8.2.2. Biological Locomotion Use Cases

To investigate the potential of ultrasonic lubrication as an active friction control method for locomotion, we examined two representative biological gaits with complementary modes of locomotion. The first is inchworm locomotion, which operates through sequential anchoring and advancement of body segments. The second is peristaltic slider locomotion, exemplified by the wasp ovipositor, which relies on the reciprocating sliding of multiple valves. These two use cases were selected because they have inspired numerous robotic designs, making them well-suited as benchmarks for proof-of-concept demonstrations.

The inchworm achieves locomotion through discrete phases of anchoring and advancing. It alternately grips the substrate with its front and rear prolegs, contracts its body to bring the rear forward, then extends again to project the front forward. At each stage of the cycle, one part of the body remains fixed while the other is moving, ensuring sequential progress. This stop-and-go strategy produces net displacement without requiring continuous wave propagation and has inspired a range of robotic implementations, often termed “inchworm robots” or “two-anchor crawlers,” where sequential anchoring is achieved with frictional pads, clamps, or suction elements.

Parasitic wasps achieve locomotion of their ovipositor through a peristaltic slider mechanism, where three slender valves—one dorsal and two ventral—are held together by a tongue-and-groove connection that constrains them radially while permitting longitudinal sliding. During penetration, locomotion is generated by sequential reciprocation: while one valve advances forward into the substrate, the other two remain stationary and provide anchoring through friction, counteracting the penetrating force. The advancing valve then becomes stationary while another valve moves forward, and this cycle repeats across all three valves. After each has advanced in

turn, the valves reset in unison, producing a net forward displacement of the ovipositor. By iterating this cycle, the ovipositor is able to steadily penetrate deep into host tissue or other substrates with minimal overall insertion force.

## 8.3. Design and Implementation Ultrasonically-Lubricated Bio-inspired Locomotion Systems

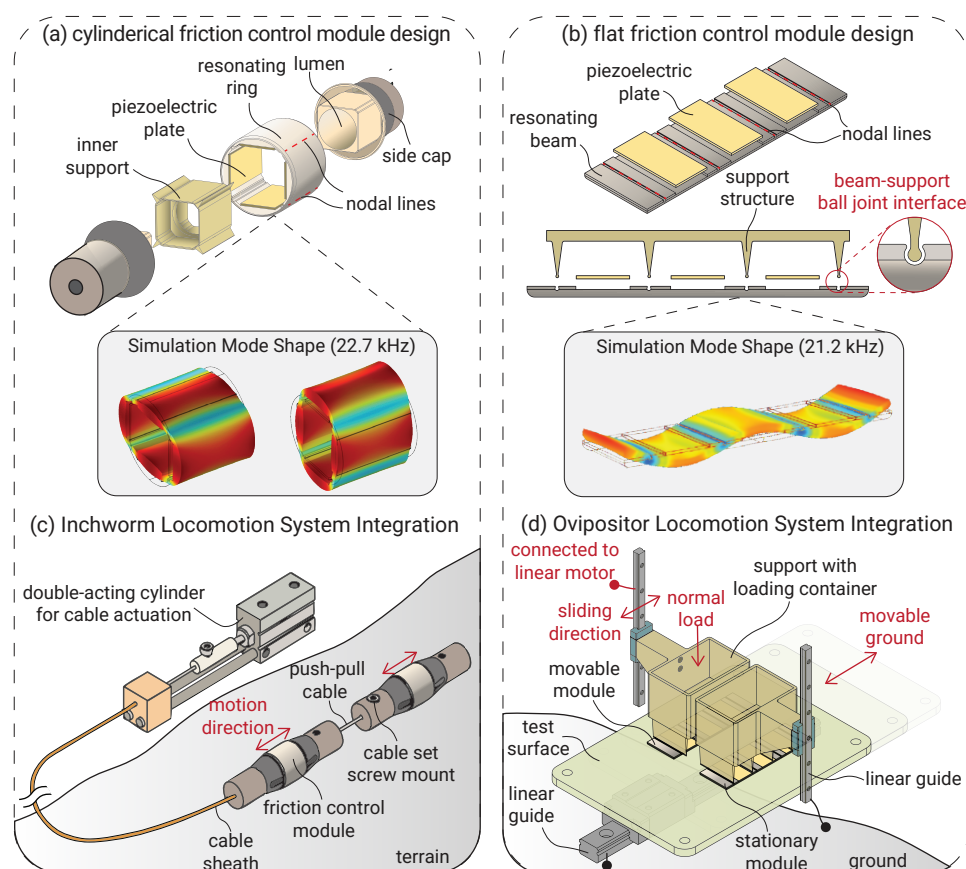
### 8.3.1. Design Requirements

To implement the two locomotion strategies using ultrasonic lubrication, the first step is to develop a friction control module, which is the fundamental building block of the locomotion system. We present two module designs: a cylindrical block intended for locomotion within lumens (internal locomotion) and a flat-plate block designed for locomotion on external surfaces. For ultrasonic lubrication to be effective, each module must be able to produce a vibration amplitude of  $\geq 20$  kHz at an ultrasonic frequency of  $\geq 20$  kHz [32, 33].

Both modules share a common design principle. The resonating structure is engineered with a flexural resonance mode in the ultrasonic range. Piezoelectric plates are positioned at the antinodes of this mode so that, when activated, they excite the structure at its designed flexural resonance mode. To allow the structure to vibrate freely, structural supports are positioned at the nodal lines of the resonance mode, where the displacement is nearly zero. These supports, implemented through grooves and needle-like contact elements, allow free rotation at the nodes, and thus reduce interference with vibration. This approach enables the resonating structure to act as a mechanical amplifier, achieving the vibration amplitudes required for effective ultrasonic lubrication (Fig. 8.2(a)(b)).

### 8.3.2. Cylindrical Friction Control Module Design

The cylindrical module consists of a ring-shaped resonator with four bonded piezoelectric plates, designed to operate in its second flexural resonance mode, as illustrated in Fig. 8.2(a). The prototype used for characterization had an outer diameter of 10 mm, following the design of [31]. The ring is supported internally by a frame with needle-like edges contacting the resonator at nodal lines. Finite element analysis, using stainless steel as the material, predicted a resonance frequency of 22.7 kHz and a maximum vibration amplitude of approximately  $4 \mu\text{m}$  per 100 V excitation.



**Figure 8.2. Design and integration of cylindrical and flat friction control modules for bio-inspired locomotion.** (a) Cylindrical module: a 10 mm ring-shaped resonator with four bonded piezoelectric plates, supported at nodal lines by an internal frame to minimize interference with oscillation. Finite element simulation predicted the second flexural resonance mode at 22.7 kHz. (b) Flat module: a  $32 \times 10 \times 1$  mm slider resonator with piezoelectric plates bonded at antinodes and supported via a ball-joint groove at nodal lines. Simulation results predicted the third flexural resonance mode at 21.2 kHz. (c) Inchworm-inspired locomotion system: two cylindrical modules connected by a push–pull cable-sheath mechanism and actuated by a double-acting cylinder, enabling bidirectional motion through selective activation of each module. (d) Ovipositor-inspired locomotion system: two flat modules, one fixed and one mounted to a linear actuator, enabling bidirectional motion through selective activation of the movable module during the motion cycle.

### 8.3.3. Flat Friction Control Module Design

The flat module is a rectangular slider resonator ( $32 \times 10 \times 1$  mm) with three piezoelectric plates bonded at the antinodes, designed to operate in its third flexural mode, as shown in Fig. 8.2(b). The slider is supported at its nodal lines through small circular cutouts that accommodate a ball-joint groove and thin-legged supports, constrain-

ing the structure while allowing free rotation. Finite element analysis, using stainless steel as the material, predicted a resonance frequency of 21.2 kHz and a maximum vibration amplitude of approximately  $4.73 \mu\text{m}$ .

#### 8.3.4. Prototype Development

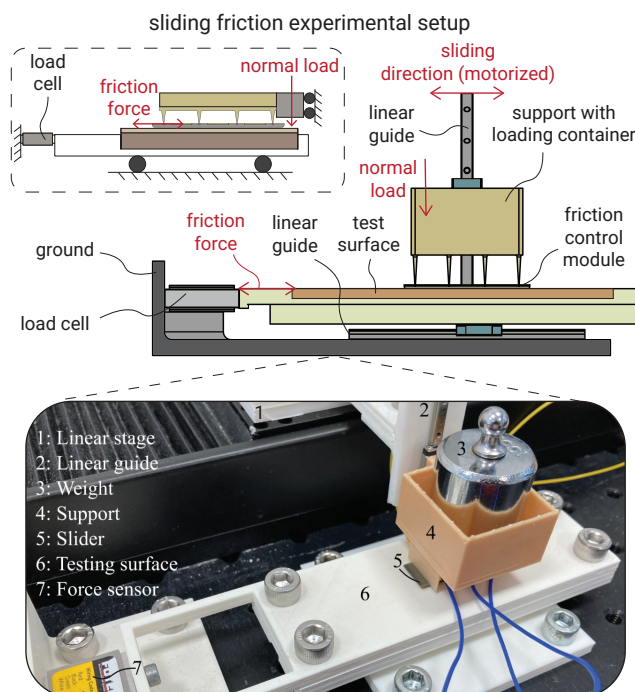
Friction control modules, both flat and cylindrical, were fabricated from stainless steel 316 using electrical discharge machining (EDM). Piezoelectric plates (SM112, Steminc, FL, USA) were cut to size and bonded into pre-designed slots with a two-component, electrically insulating adhesive (3M Scotch-Weld Epoxy Structural Adhesive DP490). Each plate included two electrodes on one side, with the back electrode bonded to the resonator surface. Wires were soldered to the exposed electrodes and arranged so that consecutive plates were driven with a phase shift of  $180^\circ$ , ensuring excitation of the target resonance mode. The support structures were produced using a stereolithography (SLA) 3D printing (Formlabs, USA), providing the precision required for the fine support legs interfacing with the resonating structures.

#### 8.3.5. Locomotion System Integration

The inchworm-inspired locomotion system was realized using two cylindrical friction control modules with an extensor element positioned between them, as illustrated in Fig.8.2(c). The extensor was implemented through a push–pull cable-sheath mechanism, in which the sheath was anchored to the first module while the inner cable passed through and was fixed to the second module. The push–pull assembly was driven by a double-acting linear cylinder that actuated the extension and contraction. A computer-programmed control system coordinated the actuation sequence by selectively activating each friction control module based on the desired direction of motion and the phase of the motion cycle.

The ovipositor-inspired locomotion system was constructed using two flat friction control modules, as illustrated in Fig.8.2(d). One module was fixed in place and the other was mounted to a linear actuator that provided reciprocating motion while the ground was mounted on a linear bearing to slide freely. A computer-programmed control system was then used to selectively switch the vibrations of the moving slider on or off, depending on the desired direction of motion and the phase of the motion cycle. By fixing one slider and actuating the other, while allowing the contacting ground surface to slide freely, the same principle of ovipositor locomotion could be





**Figure 8.3. Experimental setup for the friction characterization experiments.** A miniature load cell mounted along the sliding axis measured friction forces via a bracket guided on a horizontal stage, while the slider was mounted to a vertical linear guide for motorized horizontal motion and free vertical motion. A 100 g weight ( $\approx 1$  N) provided the normal load. Two protocols were tested: (i) sliding with and without continuous vibration at a voltage amplitude of 140 V to assess stability of the lubrication effect when activated, and (ii) sliding with the excitation voltage amplitude ramped up from 0 to 140 V to evaluate dynamic friction modulation. Tests were performed on dry and wet PLA, sandpapers of different grit sizes, granular soil, and ex-vivo porcine intestinal tissue.

demonstrated in a simpler and functionally equivalent manner that is better suited for experimental evaluation.

## 8.4. Experimental Validation

To validate the concept of the friction control modules and characterize their performance, we conducted three experiments. The first experiment aimed to characterize and validate the vibration performance of individual modules. The second was a proof-of-concept experiment to demonstrate locomotion enabled by ultrasonic lubrication in both bio-inspired designs and to quantify the resulting locomotion efficiency. The third aimed to assess the ability of the modules to modulate friction across surfaces with different materials and conditions.

### 8.4.1. Vibration Characterization Experiment

The resonance properties of the friction control modules were evaluated by measuring the resonance frequency and the amplitude of the vibration through two complementary tests. First, a frequency response measurement was used to identify the resonant frequency. Second, a voltage sweep was performed at this frequency to quantify the vibration amplitude as a function of the input voltage. Both unsupported and supported sliders were tested to assess the influence of the support structure, and each experiment was repeated six times.

The vibration response was measured using a laser Doppler vibrometer (Polytec OFV-5000 Vibrometer Controller). The Laser Doppler Vibrometer tracked the velocity of a marked anti-nodal point at the center of the slider, while the prototypes were powered by a piezo amplifier (PiezoDrive PD200X4 Voltage Amplifier). The Laser Doppler Vibrometer and amplifier were connected to a data acquisition system (National Instruments USB X Series Multifunction DAQ), which interfaced with a PC for control and data logging.

For the frequency response test, the modules were activated, and the excitation frequency was swept from 20 to 24 kHz at maximum voltage to determine the resonant frequency. For the amplitude test, the frequency was fixed at the identified resonance frequency, and the input voltage was increased from 0 to 100 V in steps of 10 V to record the corresponding vibration amplitudes.

### 8.4.2. Proof-of-concept Locomotion Experiment

We tested both locomotion systems on rigid 3D-printed PLA substrates to demonstrate the viability of ultrasonic lubrication for locomotion and to quantify locomotion efficiency. The inchworm-inspired system was evaluated on a semi-cylindrical substrate that matched the geometry of the cylindrical friction control modules, while the ovipositor-inspired system was tested on a flat substrate. Each system was first actuated without ultrasonic lubrication to establish a baseline, after which the same motion sequence was repeated with the ultrasonic lubrication activated. For each test, the distance traveled by the system  $d_{actual}$  was measured and compared to the actuator stroke  $d_{theoretical}$ —the theoretical displacement the locomotion system should achieve without slipping—to determine the locomotion efficiency as follows:

$$\eta = \frac{|d_{actual} - d_{theoretical}|}{d_{theoretical}} \times 100 \quad (8.1)$$

### 8.4.3. Friction Characterization Experiment

Following the proof-of-concept, we conducted sliding friction experiments using only the flat friction control module to evaluate the slider's ability to modulate friction across a range of surface conditions: a rigid 3D-printed PLA surface under both dry and wet conditions, a granular soil surface, and surfaces with varying roughness using coarse (150 grit) and fine (240 grit) sandpapers. Finally, we performed tests on wet ex-vivo biological tissue (porcine intestinal tissue) to explore the viability for medical applications.

Friction forces were measured with a miniature load cell (Futek LSB200) mounted along the sliding axis and connected to the test surface through a mounting bracket, which was supported by a horizontal linear guide (IKO Nippon Thompson LWLC3C1R60T0H, LWL) to allow free motion in the direction of the friction force, as illustrated in Fig.8.3. The slider was attached to a linear actuator (Thorlabs NRT150/M) through a vertical linear guide, providing controlled horizontal displacement while allowing free vertical motion of the slider to apply the normal load using weights.

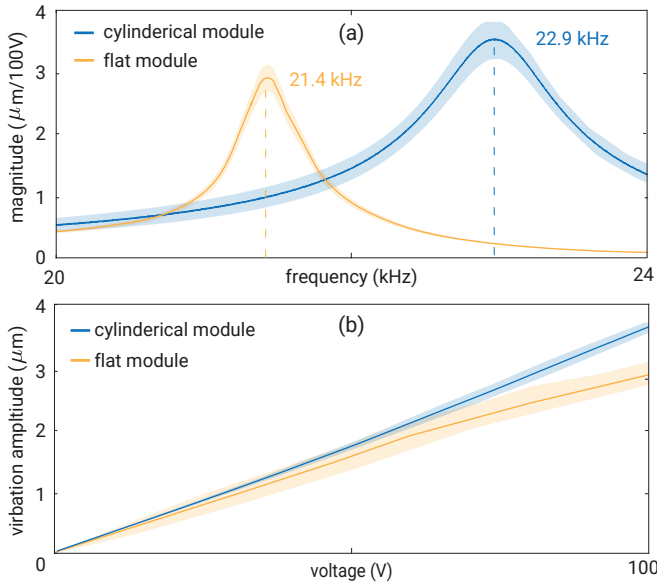
In this experiment, two sets of tests were conducted. In the first, the friction control module was loaded with a 100 g weight ( $\approx 1$  N normal load) and driven forward and backward over a 10 mm distance at 1 mm/s. The test was performed first without vibrations and then with continuous activation at a voltage amplitude of 140 V to evaluate the stability of the lubrication effect under steady excitation. In the second set of tests, the module's ability to modulate friction was assessed by sliding it over a 5 mm distance at 1 mm/s under the same 1 N load while gradually ramping the input voltage amplitude from 0 to 140 V. To quantify the module's friction modulation capability, we used the friction reduction percentage, which was calculated as follows:

$$\text{friction reduction (\%)} = \frac{\mu_{\text{off}} - \mu_{\text{on}}}{\mu_{\text{off}}} \quad (8.2)$$

where  $\mu_{\text{off}}$  is the average coefficient of friction in the non-lubricated 'off' case, and  $\mu_{\text{on}}$  is the coefficient of friction in the lubricated 'on' case.

## 8.5. Results and Discussion

The vibration characterization experiments revealed the resonance properties of both types of friction control modules. The cylindrical ring prototype resonated at 22.9 kHz,



**Figure 8.4. Vibration characterization of cylindrical and flat friction control modules.** (a) Frequency response showing resonance peaks at 22.9 kHz for the cylindrical module and 21.4 kHz for the flat module. (b) Vibration amplitude as a function of applied voltage, exhibiting linear growth up to an amplitude of  $\approx 4 \mu\text{m}$  at 100 V. Both modules achieved vibration amplitudes exceeding  $2 \mu\text{m}$  at ultrasonic frequencies ( $\geq 20 \text{ kHz}$ ), fulfilling the design requirements of ultrasonic lubrication.

at which it produced a vibration amplitude of  $\approx 3.7 \mu\text{m}/100\text{V}$ . The flat module prototype resonated at 21.4 kHz (Fig. 8.4(a)) with an amplitude of  $\approx 3 \mu\text{m}/100\text{V}$ . In both cases, the vibration amplitude increased approximately linearly with the input voltage (Fig. 8.4(b)) and reached  $\approx 4 \mu\text{m}$  before saturating due to the Laser Doppler Vibrometer measurement limit, indicating that at higher voltages, the vibration amplitude can be greater. These results confirm that both modules meet the ultrasonic lubrication design requirements, producing amplitudes greater than  $2 \mu\text{m}$  at ultrasonic frequencies ( $\geq 20 \text{ kHz}$ ).

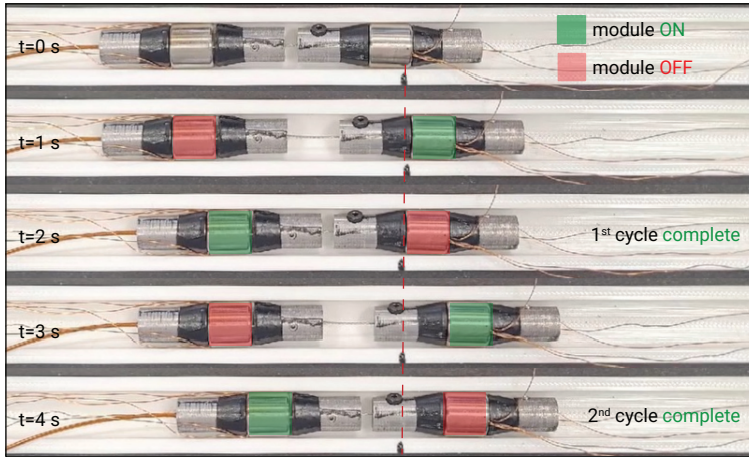
In the proof-of-concept locomotion experiments, both systems achieved locomotion only when ultrasonic lubrication was activated (Fig. 8.5) and failed to move when it was inactive. Without ultrasonic lubrication, the two friction-control modules experienced symmetric friction and therefore generated no net traction over a cycle. Activating ultrasonic lubrication selectively reduced friction at one module, allowing it to slide while the other remained stationary due to its high friction state. This selective friction control broke the frictional symmetry of the modules, creating the asymmetry required for locomotion. In both systems, bi-directional locomotion was obtained

simply by reversing the ultrasonic lubrication activation sequence on the modules (see supplementary videos 1 & 2). The inchworm-inspired system achieved a mean locomotion efficiency of 94.75%, and the ovipositor-inspired system achieved 93.2%. These locomotion efficiencies are specific to the experimental setup and normal loading conditions; however, they demonstrate the potential of ultrasonic lubrication as a friction-control mechanism to reliably achieve locomotion and enhance its efficiency.

The sliding friction experiments demonstrated the viability of ultrasonic lubrication across a range of surfaces (Fig. 8.6), achieving average friction reductions of 71–77% on dry PLA,  $\approx$  56%–71% on wet PLA, 28–49% on soil and sandpaper, and  $\approx$  27–38% on ex-vivo biological tissue. In the steady-sliding experiments (Fig. 8.6(a)), the flat friction-control module was first slid forward and backward on each surface without activating ultrasonic lubrication to establish the baseline friction forces. The same forward–backward sliding cycles were then repeated with the ultrasonic lubrication activated. In all cases, activating ultrasonic lubrication reduced the measured friction forces compared to the baseline case, and the reduction remained stable over the forward and backward cycles. This shows that ultrasonic lubrication can provide a steady and consistent direction-independent reduction in friction, rather than a transient effect.

The friction modulation experiments (Fig. 8.6(b)) further showed that the ultrasonic lubrication effect is controllable beyond a simple on–off state. In these experiments, the input voltage (and thus the vibration amplitude) was varied to modulate the coefficient of friction. As shown in Fig. 8.6(b), changing the voltage yielded a well-defined, continuous range of friction coefficients across all tested surfaces, demonstrating that ultrasonic lubrication can be used to finely tune friction simply by adjusting the vibration level.

The friction reduction levels and trends across the different substrates help explain the underlying differences in performance, as summarized in Fig. 8.6(c) for both the steady-sliding comparison (left) and the voltage-modulation case (right). On smooth dry PLA, the friction reduction is consistent with the formation of an air squeeze film. Because air is compressible, the film thickness develops nonlinearly with excitation and eventually saturates, which matches the nonlinear trend observed in the voltage modulation experiments (Fig. 8.6(b)). On wet PLA, ultrasonic lubrication was still able to reduce friction via the formation of a pressurized liquid squeeze film. Unlike air, liquids are nearly incompressible, and vibrations generate squeeze-film pressure



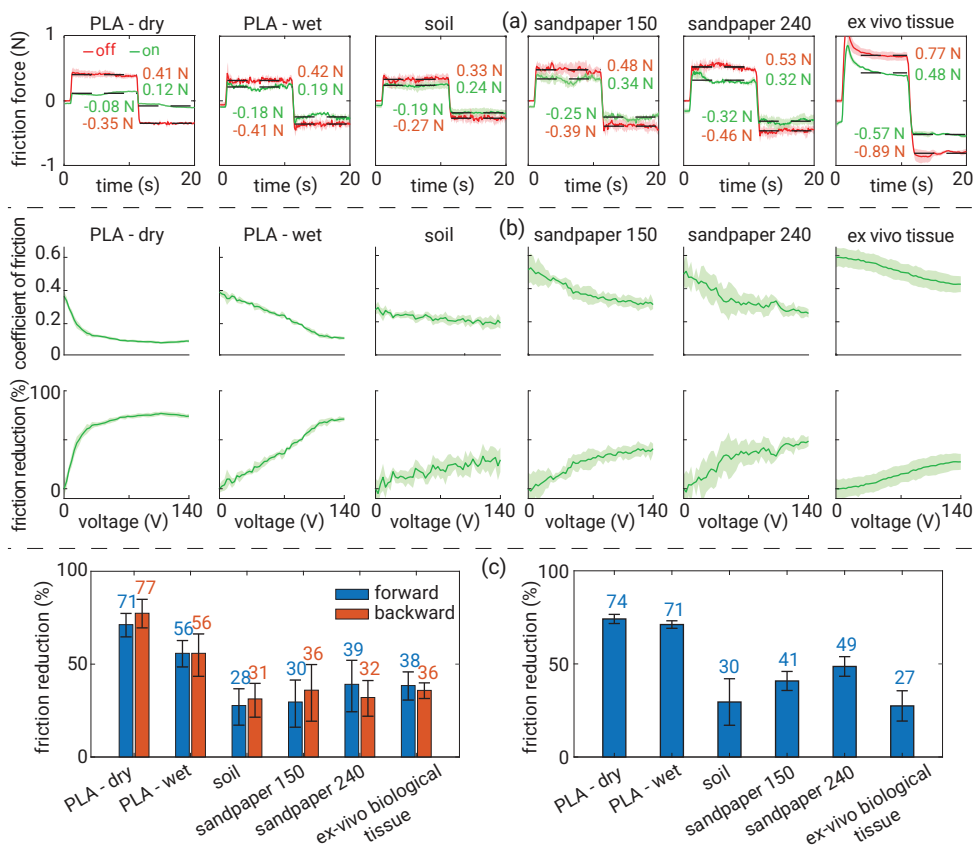
**Figure 8.5. Time-lapse sequence of the inchworm locomotion system traversing a rigid PLA track.** By selectively activating ultrasonic lubrication at each friction control module, forward propulsion is achieved following the inchworm motion cycle. Images are shown at one-second intervals. Refer to supplementary videos 1 & 2 for demonstrations of the use cases presented in this chapter, including the inchworm locomotion case.

mainly due to liquid inertia [30], leading to a more linear voltage–friction response compared to air, as shown in Fig. 8.6(b).

When tested on rough surfaces, the degree of reduction depended on surface roughness. On sandpaper, the finer grit (240) produced slightly greater friction reduction than the coarser grit (150), as the smaller asperities and voids in the finer grit case help promote stronger squeeze-film formation, resulting in an improved lubrication effect. Soil yielded relatively low friction reduction, which can be explained by its granular structure, which creates a constantly changing and non-uniform contact interface, disrupting stable squeeze-film formation, and thus reducing the effect of ultrasonic lubrication. Finally, the ex-vivo porcine intestinal tissue interface showed a reduced effect of ultrasonic lubrication compared to that of the rigid wet PLA case, which can be attributed to the viscoelastic properties of tissue. The viscoelasticity of the tissue dissipates part of the vibration energy, which limits the pressure buildup in the squeeze film and thus reduces the lubrication effect.

## 8.6. Conclusion

We introduced ultrasonic lubrication as a friction-control mechanism for robotic locomotion. We presented two friction control module designs: cylindrical and flat, and in-



**Figure 8.6. Friction modulation experiments across a range of surfaces.** (a) Measured friction forces under forward and backward sliding show that the vibrating slider effectively reduces friction on PLA (dry and wet), soil, sandpaper, and biological tissue (colon). (b) Voltage modulation of the coefficient of friction (top row) and corresponding friction reduction percentage (bottom row). On PLA, reduction is consistent with air squeeze-film formation and exhibits nonlinear saturation. On wet PLA, ultrasonic lubrication enabled friction reduction via liquid squeeze-film formation, following a more linear trend compared to air. Sandpaper surfaces show higher reduction for finer grit (240) compared to coarser grit (150), due to enhanced squeeze-film effects in smaller asperities. Soil produced relatively low friction reduction owing to its unstable granular contact. On colon tissue, viscoelasticity damped vibration energy, decreasing friction reduction. (c) Summary of friction reduction percentage on different substrates during experiments on forward–backward sliding at maximum voltage (left) and experiments with voltage modulation (right).

tegrated them into two locomotion systems inspired by inchworm and parasitic wasp ovipositor motion. Vibrometry experiments confirmed that both modules can generate amplitudes exceeding  $2 \mu\text{m}$  at ultrasonic frequencies around 22 kHz, fulfilling the requirements of ultrasonic lubrication. In proof-of-concept locomotion experiments, ultrasonic lubrication enabled bi-directional locomotion in both inchworm-inspired and ovipositor-inspired systems, with mean efficiencies of 94.75% and 93.2%, while

no locomotion was achieved when ultrasonic lubrication was not activated due to friction symmetry. Sliding friction experiments demonstrated the ability of ultrasonic lubrication to achieve stable and tunable friction reduction across a variety of rigid and soft interfaces in dry and wet conditions, highlighting the versatility of ultrasonic lubrication for robotic locomotion applications.

In future work, we will focus on modeling the dynamics of ultrasonic lubrication in locomotion, with particular attention to activation timing during the gait cycle. By relating activation phases to locomotion efficiency and stability, we aim to develop optimized friction control strategies for locomotion and explore additional functionalities, such as adaptive steering and programmable interaction with complex environments.

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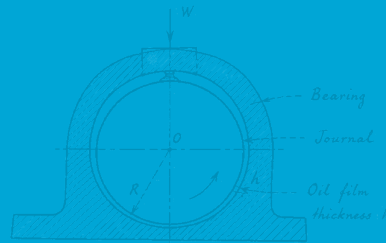
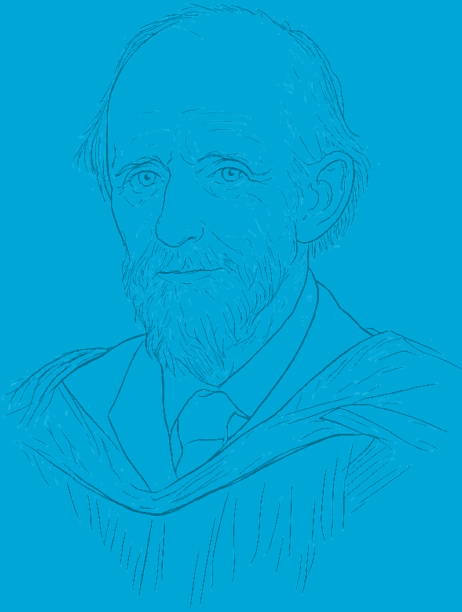


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# 9



*Lubrication, or the action of oils and other viscous fluids to diminish friction and wear between solid surfaces depends upon the surfaces being continuously separated by a film of the lubricant.*

*The subject is of such fundamental importance in practical mechanics, and the opportunities for observation are so frequent . . .*

*The subject is of such fundamental importance in practical mechanics, and the opportunities for observation are so frequent . . .*

*Osborne Reynolds.*

## Discussion

## 9.1. key Findings of This dissertation

Throughout this dissertation, we explored how we can control friction to engineer instruments that glide when low resistance is needed, grip when stability is required, and propel when motion is desired. We applied these techniques to medical instruments such as catheters and robotic capsules for endoscopy. In these applications, friction at the tool–tissue interface is both a constraint and a design opportunity: (1) it can restrict motion and increase patient risk, yet, if actively controlled, (2) it can be used to achieve safer navigation, more stable intervention, and (3) unlock new forms of locomotion.

This dissertation examined two methods for friction modulation to enhance (glide or grip) or enable (propel) function in medical instruments, followed by a final study that combined them:

- **Modulate Friction to Glide or Grip (Part I):** This part investigated the use of ultrasonic lubrication to modulate the frictional properties of endoluminal instruments, such as catheters and endoscopes, on demand, enabling them to glide for safer and smoother navigation or grip for stability.
- **Modulate Friction to Propel (Part II):** This part explored coordinated motion patterns across an array of sliding contact elements to dynamically tune friction anisotropy at the interface of self-propelling robotic capsules, enabling them to self-propel through endoluminal environments.
- **Glide and Grip to Propel (Part III):** This part explored combining local-level friction modulation (from Part I) with system-level tuning of friction anisotropy (from Part II), unlocking new possibilities in robotic locomotion.

The contributions of this dissertation addressed theoretical and foundational problems in the modulation of friction, which were key to the applications of medical instruments described in this dissertation, of which I highlight the following key findings:

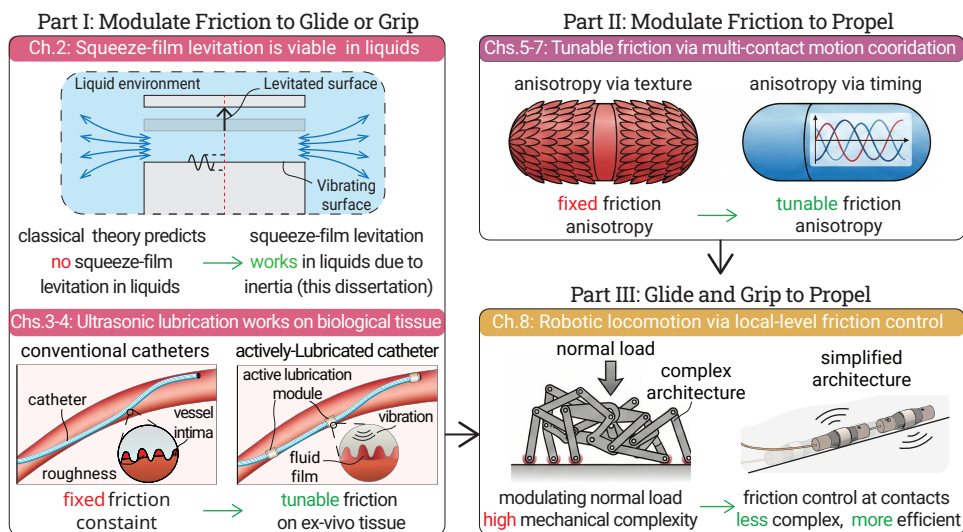
In Part I, Chapter 2 demonstrated the viability of ultrasonic lubrication to operate in submerged liquid environments (Fig. 9.1(a)), satisfying a critical requirement for endoluminal medical applications. While this technology is well-established in dry settings (e.g., haptics, industrial bearings), its viability in bodily liquids was still unproven. Classical theory links the lubrication effect to air compressibility, a property nearly absent in liquids. We derived a closed-form model to capture the physics of

incompressible squeeze films under ultrasonic transverse vibrations. The model, validated by experiments, confirmed that liquid squeeze films can carry loads under transverse ultrasonic excitation, and uncovered the physical mechanism behind the lubrication effect. This model provides a critical design tool by explicitly linking the interfacial separation between the two surfaces, which drives lubrication, to vibration amplitude and frequency, fluid density, contact area, and normal load. The proven load-carrying capacity of incompressible squeeze films shows that ultrasonic lubrication can be generated and sustained in liquids, confirming its viability for medical endoluminal applications such as catheterization.

In Chapters 3-4, we translated the established physics of ultrasonic lubrication in liquids into functional cylindrical friction-control modules for catheters and demonstrated its efficacy on biological tissue (Fig. 9.1(b)). Most significantly, we provided the first experimental evidence of ultrasonic lubrication on *ex-vivo* porcine aorta tissue, achieving up to 42% friction reduction, confirming the viability of ultrasonic lubrication. We evaluated clinically relevant interface conditions, including rigid versus soft contact (e.g., calcified versus healthy tissue) and curved geometries. Under all conditions, the modules showed robust lubrication performance comparable to established benchmark applications such as surface haptics. Finally, we showed that the thermal footprint of the modules remained within safe operating limits, further highlighting the potential viability of the concept for clinical applications.

In Part II, Chapters 5–7 demonstrated how system-level friction anisotropy can be tuned to achieve locomotion by sequencing the motion of multiple sliding contact elements, inspired by how parasitic wasps sequence the motion of their ovipositor valves (sliders) to penetrate, steer, or transport eggs without external forces (Fig.9.1(c)). We developed self-propelling robotic locomotion systems that mechanically encode the bio-inspired motion sequence to achieve locomotion. We demonstrated this principle in two designs; a flexible self-propelling locomotion system for pipeline inspection (Chapter 6), and a robotic self-propelling capsule for colonoscopy (Chapter 7), highlighting the versatility of the concept for a wide variety of applications. Moreover, we developed a mathematical model, validated experimentally, that links thrust generation to the motion sequence and pattern of the slider system, providing a general tool to design and simulate this class of self-propelling robotic locomotion systems.

In Part III, Chapter 8 coupled local-level friction modulation (Part I) with system-level friction anisotropy tuning through motion coordination (Part II), uncovering new pos-



**Figure 9.1. Summary of key findings and dissertation structure.** **Part I (Modulate Friction to Glide or Grip):** Squeeze-film levitation, classically predicted to fail in incompressible fluids, was theoretically and experimentally confirmed in liquid environments, driven by fluid inertia (Ch. 2). This phenomenon enables actively lubricated instruments to achieve a tunable coefficient of friction on *ex vivo* biological tissue, replacing the fixed friction constraints of conventional catheters (Chs. 3–4). **Part II (Modulate Friction to Propel):** Precise motion coordination and timing of multiple contacting segments enables programmable friction anisotropy, offering a dynamic alternative to static, texture-based anisotropy (Chs. 5–7). **Part III (Glide and Grip to Propel):** Direct modulation of the friction coefficient at local contacts enables efficient robotic locomotion, significantly reducing the mechanical complexity required by traditional architectures that rely on normal-load modulation (Ch. 8).

sibilities for robotic locomotion (Fig.9.1(d)). In order to assess the potential benefit of this coupling on locomotion efficiency and design, we used ultrasonic friction control modules as sliders to construct two bio-inspired systems that execute the motion patterns of the inchworm and the wasp ovipositor. The integration of local-level active friction control via these ultrasonic modules proved effective, as it enabled locomotion with minimal number of components in both bio-inspired systems. In particular, we achieved both the inchworm and ovipositor locomotion patterns with only two modules under same normal load and isotropic surface conditions, whereas nature requires either varying the normal load, surface anisotropy or using at least three sliders to move. This demonstration highlights the potential of this concept to reduce design complexity and enhance the efficiency of robotic locomotion systems.

## 9.2. Interpretation and Implications

### Incompressible Squeeze-Film Levitation: Inertia Matters

Historically, squeeze-film levitation, the driving phenomenon behind ultrasonic lubrication, was modeled almost exclusively through the lens of classical gas lubrication theory, where the load-carrying capacity is attributed to the non-linear compressibility of the air film [1]. As such, classical theory predicts the levitation effect to vanish in incompressible fluids (i.e., liquids). This prediction fundamentally contradicts the experimental observations of underwater squeeze-film levitation [2], indicating a gap in the theory. While this underwater effect was hypothesized to be driven by viscosity [2], no mathematical model was developed to support this hypothesis. Moreover, recent models predicted incompressible squeeze films to generate weak attractive forces under transverse vibrations [3], deepening the disconnect between theory and practice. To address this gap, we relaxed a core assumption of classical lubrication theory by incorporating fluid inertia—a factor typically assumed negligible in gas-based models. Our derivation of the inertial Reynolds equation confirmed that incompressible squeeze-films can carry loads, in line with the experimental observations, and therefore can generate stable levitation under ultrasonic excitation. Crucially, the model uncovered fluid inertia as the fundamental mechanism driving this load-carrying capacity, which explains why classical models, that typically assumed inertial effects to be negligible, failed to capture the effect in liquids.

Beyond this dissertation, uncovering the physics of squeeze-film levitation in liquids enables the use of ultrasonic lubrication in applications where it was previously assumed to be ineffective or unpredictable. These applications include the medical use cases explored here, as well as contactless handling of delicate, contamination-sensitive samples—such as biological specimens—inside liquid environments [4], in addition to contactless transportation and liquid squeeze-film bearing in submerged machinery [5]. Additionally, these findings are promising for acoustic manipulation of micro-scale objects such as microelectromechanical systems (MEMS) in liquid environments where overcoming viscous drag is highly needed [6, 7].

### Active lubrication: a new dimension to instrument design

Prior to this work, friction management in transluminal medical instruments, such as catheters, was largely addressed through passive surface design and procedural workarounds. Common strategies include applying liquid lubricants prior to insertion



[8–12] or using self-lubricating coatings such as hydrophilic layers that become slippery when exposed to wet surfaces [13, 14]. While effective in specific settings, these approaches are limited in durability and versatility: add-on lubricants are often short-lived as they are wiped away during navigation [8], and coatings cannot be adjusted during use and provide no active mechanism to transition between low friction for insertion and higher friction for stability. In parallel, the medical robotics literature has pursued related goals, stability, force capability, and safety, through variable-stiffness mechanisms [15] and shape-control [16], which control the body mechanics of the instrument to improve mechanical interaction and stability, however, these approaches typically assumed fixed friction properties, leaving behind the interaction challenges specifically associated with friction.

Active lubrication adds a new dimension to the design of these instruments by turning friction at the tool–tissue interface from a fixed constraint into a tunable operating variable. Beyond simply reducing friction, it enables instruments to regulate how strongly they adhere to tissue during a procedure on demand, creating new possibilities for safer navigation, stable positioning, and controlled force delivery. The broader implication is a shift in design perspective from choosing a single “best” surface to actively *scheduling* and *modulating* friction during operation. Ultrasonic lubrication can be tuned continuously between high and low friction levels, allowing instruments to target an intermediate resistance when needed rather than only switching between extreme states. Moreover, if friction is controlled independently across multiple segments, it can enable richer functionalities beyond glide-or-grip behavior, such as steering and bending by creating temporary anchors at selected locations. More generally, active friction modulation can complement variable-stiffness approaches: stiffness can provide internal load-bearing when required, while friction modulation can manage external coupling to tissue.

## Designing friction anisotropy through timing, not texture

Prior to this work, a number of bio-inspired self-propelling systems have demonstrated the concept of motion coordination to achieve locomotion, often relying on multi-segment peristaltic or inchworm-like gaits [17, 18]. However, the influence of motion patterns on thrust generation was often treated implicitly. The modeling framework presented in this dissertation connects the ovipositor-inspired motion sequencing of multiple sliding contacts directly to thrust generation, providing a model-based method to tune friction anisotropy primarily through timing rather than relying

on texture. In this paradigm, desired frictional anisotropy behavior can be encoded into the mechanical structure, potentially reducing the need for sophisticated control and redundant actuation mechanisms, which establish a pathway toward embodied mechanical autonomy in self-propelling systems.

Looking forward, active tuning of friction anisotropy opens compelling new avenues for adaptive medical robotics. Because thrust generation is governed by the motion sequence rather than fixed physical textures, future devices could incorporate smart mechanical transmissions or passive compliant linkages that naturally alter the timing cycle in response to varying tissue stiffness or luminal resistance. This would allow a single, mechanically simple capsule to autonomously shift its “gait” when transitioning from a highly compliant environment to a highly constrained one, realizing true embodied mechanical adaptation without increasing electronic or computational complexity.

## Designing with friction: A new paradigm for locomotion

In nature and bio-inspired robotics, multi-contact locomotion systems, such as inchworm- or earthworm-inspired gaits, typically propel by creating a friction differential between contacts. They do this by modulating the normal load at individual contacts (clamping/unclamping, adhesion) and/or using directional surface features (e.g. scales) that are coupled with a motion sequence of the contacts during a cycle [19–23]. While effective, these strategies often add design overhead: normal-load modulation requires extra hardware and control, and surface-texture anisotropy requires advanced fabrication techniques and offers little on-demand tunability, both of which complicate the system design and control [19, 21–23]. This dissertation introduced a different control handle for friction anisotropy in multi-contact locomotion systems: direct modulation of the coefficient of friction ( $\mu$ ) at the interface via ultrasonic lubrication [24, 25]. By actively modulating  $\mu$  at individual contacts, their contact state can be regulated between ‘grip’ and ‘slip’ independent of normal load and surface texture, improving controllability while reducing design overhead, as demonstrated by the inchworm and ovipositor gait use cases in Ch. 8.

This paradigm shifts the focus of the literature from designing locomotion systems ‘around friction’ to designing ‘with friction’. Rather than increasing the number of contacts or amplifying normal forces to generate traction, locomotion is achieved by directly modulating the coefficient of friction between contacts independent of the

normal load and the surface design. By decoupling the friction state (anchor or release), this paradigm enables simpler and highly adaptable locomotion architectures capable of tuning their frictional interaction with the environment on demand. Ultimately, it establishes a pathway for engineered locomotion systems to go beyond the limits of nature-inspired solutions.

### 9.3. Limitations and Open Questions

While this dissertation demonstrates the feasibility of active friction modulation and coordination-driven locomotion, several theoretical and practical limitations remain.

In Part I, the incompressible squeeze-film model provided a useful foundation for understanding levitation in liquid environments. However, it does not account for the viscoelastic response of soft tissue at the interface. Since biological tissues exhibit rate-dependent behavior, incorporating viscoelasticity into the model could improve predictive accuracy and inform the design of vibrating elements under realistic conditions. In addition, the energy cost of levitation in liquid remains an open question. Although liquid interfaces can generate a load-carrying capacity greater than that of air, achieving the same vibration amplitude in a liquid is expected to require greater energy due to increased inertia and viscous drag. Therefore, a systematic evaluation of energy consumption versus friction reduction is necessary to assess efficiency.

From an engineering perspective, several technical challenges remain for ultrasonic lubrication in the catheter application. Reliable sealing of the vibrating friction control modules proved difficult as rigid sealing interfaces dampen vibrations and reduce performance, while flexible sealing membranes did not provide sufficient sealing to enable testing in fully-submerged settings. Matching resonance across multiple modules also presents a challenge: small fabrication inaccuracies can slightly shift the resonance frequency a module, which can result in inconsistent lubrication effect across modules when driven by a fixed frequency signal. This poses challenges for scaling the approach to multi-module instruments and for miniaturization, where higher resonance frequencies demand tighter manufacturing tolerances. Finally, the use of ultrasonic lubrication is limited by the fact that it only reduces friction. The upper limit of the coefficient of friction, and thus the system's stabilizing grip, remains passive, implying that the device must be designed with a baseline friction high enough to provide stability by default.

In Part II, locomotion concepts based on coordinated sliding elements were validated primarily in simplified and *ex-vivo* environments. The behavior of the locomotion systems in anatomically complex geometries, such as tight bends, heterogeneous compliance, and dynamic contact conditions—remains to be studied. As such, the validity of the modeling framework put forward in this dissertation requires additional validation in order to assess its accuracy in more representative conditions.

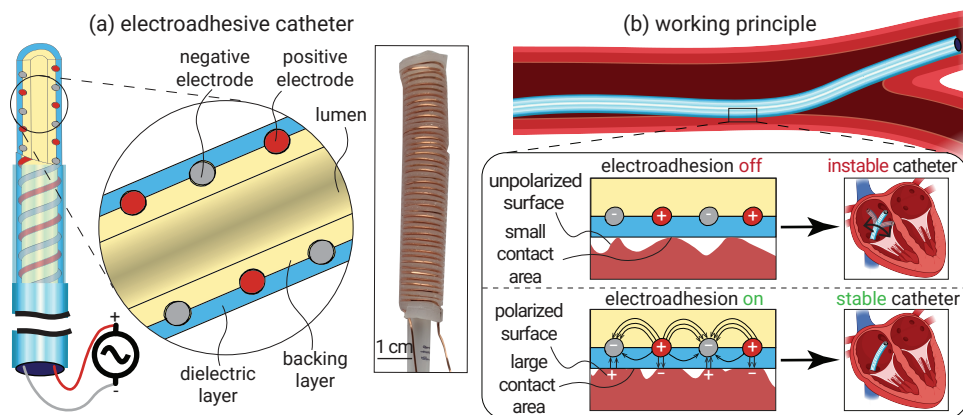
Finally, in Part III, combining local friction modulation with system-level coordination raises a key energy question: does the power needed to drive the ultrasonic elements outweigh the benefits in locomotion efficiency and reduced mechanical complexity? In other words, the energy spent to modulate friction must be compared to the energy saved by reducing sliding friction and redundant actuators. Quantifying this trade-off is essential to justify the approach as a practical route to simpler and more efficient self-propelling robots.

## 9.4. Future Outlook and Opportunities

This dissertation demonstrated that friction is not merely a passive system property, but a design variable that can be actively modulated to enhance or enable functions in medical instruments and robotic locomotion systems. Building on the methods and findings established here, several future directions follow, spanning mechanisms, modeling, and hardware design.

While this dissertation focused on using ultrasonic vibrations to reduce friction, recent studies reported that flexible disks vibrating at low frequencies ( $10^2$ – $10^3$  Hz) generate a controllable attractive squeeze-film force that enables load-carrying adhesion [28, 29]. Following this approach, future work can explore the design of active friction control modules that increase friction by generating adhesion, an inverted workflow to the one proposed in this dissertation. In this scenario, modules will be designed with a default low-friction state and can be switched to high-friction by activating vibrations. Another direction is to investigate the design of a friction control module that can switch between levitation and adhesion by tuning the vibration parameters. Achieving this dual behavior would expand the controllable friction range, which would enable more optimized 'glide' and 'grip' performance.

Another direction is to explore using vibrations not only to modulate friction, but also to estimate contact properties by tracking the dynamic response of the vibrating modules. Prior work has shown that ultrasonic vibrations could be used to infer the coef-



**Figure 9.2. Preliminary work on electroadhesive catheters** [26, 27] (a) Electroadhesive catheters concept and preliminary prototype. (b) Working principle of electroadhesive catheters.

ficient of friction at the interface [30], and to detect stick–slip transitions in vibrating frictional contacts [31]. Coupling sensing with friction modulation would enable closed-loop systems that adjust their friction state based on the detected contact condition to achieve a desired behavior.

Investigating alternative ways to deliver ultrasonic vibration to the surface of an instrument is another promising direction. Instead of integrating active resonators near the contact interface, vibrations could be activated remotely by embedding passive resonating structures with pre-designed resonance properties, and exciting them using an external wave at the right frequency. The excitation wave could potentially be transmitted through the surrounding environment—for example, using the human body as a medium in a way similar to ultrasound imaging [32]. Alternatively, excitation could be routed via structural or fluidic waveguides [33, 34], or through embedded acoustic metamaterials that can potentially propagate and localize vibration where needed [35]. Collectively, these approaches can potentially enable scalable and modular ways toward active friction control in long, slender devices such as catheters and beyond.

From a fabrication perspective, producing resonant friction-control modules with consistent frequency response is crucial for multi-module systems. One promising direction is to use additive manufacturing to fabricate monolithic modules with embedded resonant structures, reducing reliance on multi-part assembly. Recent work has demonstrated 3D-printed piezoceramic ultrasonic transducers integrated into a packaging structure for endovascular applications [36]. Moreover, recent studies

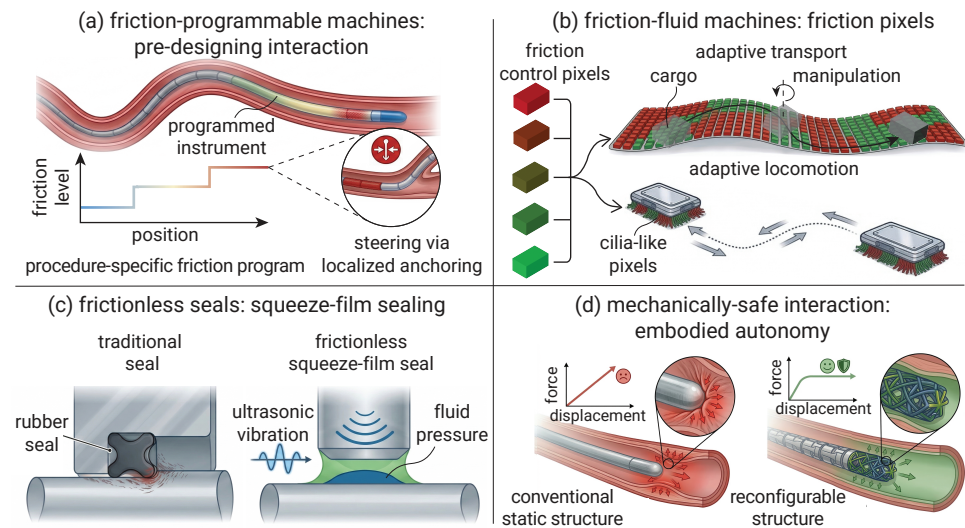
reported promising progress in non-assembly additive manufacturing of complete medical device prototypes [37]. Building on these advances, an intriguing question is whether an entire instrument can be printed with embedded friction-control modules in a single fabrication step. Such an approach could improve repeatability and scalability of fabrication and enable more optimized resonator geometries and thus better friction-control performance.

For long slender structures, such as catheters, alternative friction-modulation techniques are also worth exploring. One option is electroadhesion, where a voltage applied to embedded electrodes beneath a thin dielectric layer creates an electric field that generates electrostatic attraction to the opposing surface. These thin layers can be potentially embedded into the instrument wall via thin film coating, offering a promising scalable integration. We explored the viability of this approach in a proof-of-concept study, in which a conductive coil was embedded in a soft cylindrical actuator (Fig. 9.2) and produced measurable attraction forces in friction experiments [26, 27]. Future work can clarify how electroadhesion compares to, and potentially complements, ultrasonic friction modulation in medical contexts.

For self-propelling systems, several future directions build naturally on the locomotion concepts demonstrated in this dissertation. While this work achieved propulsion through coordinated motion sequence of discrete sliding contacts, generating a similar motion sequence on a continuous surface could improve performance and reduce design complexity. One potential option is surface traveling waves. Interestingly, ultrasonic traveling waves could combine friction modulation and locomotion in a single integrated actuation mechanism: the ultrasonic traveling wave can modulate friction through squeeze-film effects while simultaneously generating thrust force [38]. Additionally, the modeling framework, linking the ovipositor motion pattern to thrust generation, could be extended to explore the influence of general motion patterns on the sliding friction force and speed of multi-contact locomotion systems, in line with the state-of-the-art [39].

## 9.5. Visionary Outlook

The findings of this dissertation suggest a broader opportunity: once friction can be actively controlled, the interface stops being a passive boundary and becomes an active functional layer, unlocking new possibilities for future machines whose coupling to the environment can be scheduled in time and space.



**Figure 9.3. Visionary outlook of friction-modulated machines.** (a) *Friction-programmable machines*: instead of merely switching between low- and high-friction states on demand, a device could execute a predetermined, procedure-specific friction program along its trajectory, enabling customized interaction with the environment and additional functionalities such as steering via localized anchoring. (b) *Friction-fluid machines*: surfaces composed of independent friction control movable 'pixels' could spatially pattern friction, and couple this friction pattern with system-level motion sequence to realize functions such as adaptive transport, manipulation, and locomotion, analogous to artificial cilia. (c) *Frictionless seals*: ultrasonic squeeze-film pressure could be used not only to reduce friction, but also to create a pressurized fluid barrier that seals while preserving relative motion, thereby overcoming the conventional trade-off between sealing performance and friction. (d) *Mechanically safe interaction*: embedded reconfigurable compliant structures could autonomously regulate contact forces, reducing reliance on continuous sensing and control and enabling inherently safe, mechanically intelligent machines.

This dissertation demonstrated methods to control friction on demand. A natural next step is to ask whether friction can be *programmed* rather than merely controlled. If frictional interaction of a machine with a given environment can be prescribed, then this frictional interaction could be optimized and scheduled prior to its deployment, instead of being regulated reactively.

While this work demonstrated the capability to toggle between low friction for safe navigation, and high friction when stability is needed, a programmable machine would produce a desired predetermined frictional behavior/states with its environment, and can execute a pre-planned schedule of this programmed behavior, enabling customized environment-specific interaction as well as additional functionalities such as active steering and follow-the-leader shape control (Fig. 9.3(a)). In medical instruments, this would enable fully pre-planned procedure-specific interaction programs that are designed to improve clinical outcomes.

A direct extension of this programmable vision is to build friction-fluid machines composed of controllable friction 'pixels' (Fig. 9.3(b)). Instead of a machine possessing a single global friction property, the surface would be made of unit cells (pixels) whose frictional state can be set independently. Such a spatial friction 'resolution' would enable friction patterning along the surface, using contact with the environment to realize functions such as controlled manipulation through friction gradients and steering or shape control through temporary localized anchors.

In medical devices, friction pixels could enable instruments that steer and control their own shape by localized anchoring to the surrounding tissue while remaining atraumatic elsewhere, allowing for follow-the-leader motion control. At larger scales, a distributed grid of friction control pixels could act as artificial cilia. Similar to biological cilia that act in synchrony to move matter, coordinating when and where each pixel anchors or releases allows the grid to shape its overall frictional interaction. By controlling the friction state of each pixel coupled with grid-level coordinated motion patterns, the system can achieve controllable and reconfigurable transportation of matter, and adaptive locomotion.

This dissertation used ultrasonic vibrations to pressurize a squeeze film and reduce friction at the contact interface between surfaces. A visionary extension is to use the same squeeze-film pressure not only for lubrication, but for *frictionless sealing* (Fig. 9.3(c)). By maintaining a stable, pressurized fluid film, the film itself could actively block fluid leakage while simultaneously acting as a frictionless bearing [40]. This would eliminate the classical engineering compromise where achieving a tight seal inherently increases friction and restricts motion in sliding joints, rotating shafts, and access ports.

The implications extend beyond medical devices. Frictionless seals could reduce energy losses, wear, and maintenance across a wide range of fluid-handling and rotating machinery. In that sense, active squeeze-film physics could become not only a tool for friction control but a mechanism for improving efficiency and sustainability at scale.

A broader message of this dissertation is that physical interaction can be made safe through mechanical means, reducing the reliance on complex sensing and control. This raises a longer-term vision: can we build machines whose interaction with the environment is guaranteed mechanically to be safe, regardless of how an operator uses them or how uncertain the environment is?



In biological systems, managing physical interactions is a distributed process: the mechanics of the body and localized neuromuscular reflexes react autonomously to environmental disturbances and noise without requiring direct, conscious control from the brain. A similar principle could guide medical instruments and robotics. If friction and contact forces are regulated autonomously at the physical interface, the ‘brain’ of the operator (whether a human or an AI agent) will be freed to focus more on higher-level guidance and decision making, rather than being consumed in local-level compensation of disturbances and noise (Fig. 9.3(d)). In this view, friction control is not only a performance tool, but a route toward mechanically intelligent, interaction-aware machines with inherent safety and simplified autonomy.

## 9.6. Conclusion

This dissertation explored friction as an active control variable to engineer medical instruments that can glide on demand, grip as needed, and propel when desired. Across catheters and robotic capsules for endoscopy, the central message is that friction at the tool–tissue interface is not a limitation to minimize, but a controllable interaction to shape for improved safety and functionality.

Looking forward, the larger implication is a paradigm shift in the design of medical instruments and robots toward engineering ‘with’ friction, not ‘around’ it, enabling more optimal designs for physical interaction and unlocking a path toward a new class of friction-programmable instruments that are mechanically simpler yet more capable. Beyond medical applications, the methods developed in this dissertation provide a transferable toolkit for active friction control in soft robotics, human–robot interaction, and tribology, promising a future of mechanically intelligent, interaction-aware machines that adapt to the physical world with inherent safety and precision.

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# Acknowledgments

While starting to write this section, I have been thinking a lot about where to begin and what to say. It has been a long journey, full of joy, setbacks, pride, impostor syndrome, fulfillment, disappointment, and a never-ending ambition to become better and keep moving forward. It was a remarkable journey in many ways, and I hope to reflect on some aspects of it here. Before that, I would first like to thank Allah, God Almighty, for all the blessings bestowed upon me, gifts that I do not deserve, and for guiding me and filling me with the courage and goodness to continue this journey together with my family, the blessing of all blessings. I cannot be thankful enough, and I hope to continue fulfilling the purpose for which Allah created us: to build bridges between hearts and nations, to contribute to good and refrain from evil, to keep our hearts alive with compassion and love for one another, to keep our message full of peace, and to wish one another goodness and prosperity. Amen.

My PhD journey at Delft was a great privilege that I do not take for granted and for which I will remain grateful for the rest of my life. The greatest gift of this privilege was the people with whom I had the pleasure of sharing this journey: my family, friends, colleagues, and mentors. To me, people are what made the journey special and gave it meaning beyond material achievements. I believe that success is never a solitary journey, but one in which a person needs to be lucky enough to be surrounded by people who help, stimulate, and encourage them to make the journey a great success. Before writing this, I spent some time reflecting on the journey and thinking about what to say. I realized that most acknowledgments begin by thanking supervisors, colleagues, and friends, rightly so, and thank the family at the end. Eventually, I decided to change that sequence and begin with my family: my father, my mother, my father-in-law, my mother-in-law, Cherouk, my wife, our children Abdelaziz and Haroun, and our incoming baby, whom we cannot wait to welcome! A PhD is a tough, though enjoyable, journey, and my family were the behind-the-scenes warriors who set the stage for it to continue. Therefore, I wanted to begin by talking about my family, without whom I would not even have started the journey in the first place.

My journey started long before I came to Delft. In fact, it started even before I was born. It began with my parents, Abdelaziz and Fatma, two young teaching assistants at Cairo University who were trying to build a life together under difficult conditions. I

still remember their stories about how they sometimes chose to walk between their small rented apartment and work to save the bus fare, and how they worked together to afford their basic needs. By the time I was born, they had already come a long way in life. I had everything I needed, although they still had to work hard to maintain a good life for my brother and me. Things were stable, at least from our perspective. The older I grow, the more I realize what it took them to reach that point: immense sacrifices for which one can never be grateful enough. I still remember my first day at primary school, going early with my father, holding his hand, arriving first in the classroom, and having him show it to me before the day began. How lucky I was to sit in such a nice classroom, an opportunity my father had not had as a child. I still remember my mother teaching me how to study independently for exams and insisting that I not meet friends to study together during exam periods, so that I could focus and avoid distraction. I later realized how invaluable these lessons were to my academic journey. There are many more things that would require an entire book to describe, and even that might not be enough. Because of the unconditional love that parents give their children, one sometimes feels that it will last forever. It does not. The moment my father passed away in 2013 profoundly shaped my perception of life. It made me understand what true loss means, that nothing is guaranteed, and that we should appreciate every moment with our loved ones because we do not know whether we will have another chance to share a moment with them. I am blessed to still have my mother with me. She spared no effort to heal the wound left by losing my father and to help me establish my life. To my father, Abdelaziz Ali Atalla, and my mother, Fatma Rasmy, thank you for everything. Without you, none of this would have been possible. I love you. I would also like to thank my brother, Ehab, who inspired me to pursue engineering and supported me along the way, especially during the early stages of my independent life and beyond.

Alongside my parents, I also had my second parents: my father-in-law, Medhat, and my mother-in-law, Nagwa. They contributed immensely to my journey, first by raising my wife and partner in this journey, Cherouk, and then by supporting me throughout it. I still remember my mother-in-law completing all our paperwork at the Dutch embassy in Egypt, even before she knew that we would be moving to the Netherlands after our journey in the United States. My father-in-law, Medhat, always believed in me and encouraged me to keep going and growing. He even called me "Dr." long before I started my PhD. To my father-in-law, Medhat, and my mother-in-law, Nagwa, thank you for everything. Without you, none of this would have been possible. I love

you. I would also like to thank my sisters-in-law, Shaimaa, Shahenda and Nadine, for being part of this journey with us.

Throughout this journey, I was blessed to have Cherouk, my wife, by my side, and we were both blessed to have our children, Abdelaziz and Haroun, in our lives. I still remember the day we arrived in the Netherlands, 28 October 2020. We were a small family, Cherouk, our son Abdelaziz, and me, arriving in the Netherlands for the first time on a journey whose destination only God knew. We came from Egypt after moving across the Atlantic from the United States, where I completed my MS, with a short stay in Egypt in between. Little did we know about the new world we were entering, but we knew that we were excited to begin this new chapter. Our arrival remains memorable in many ways, marking the beginning of that chapter. I still remember arriving at Schiphol with nine pieces of luggage and trying to manage them all as we left the airport. I also remember the angry train conductor who kept whistling at us to hurry while loading our luggage onto the train to Rotterdam, where our accommodation was, because we were taking too long. A stranger stepped in to help when he saw me struggling to load everything quickly. The world then was different from the world today. It was the middle of lockdown: everything was closed, and the country looked empty. Despite all the uncertainty surrounding our new world, we were excited rather than worried.

As the days passed, we continued building our life together, piece by piece, feeling grateful for every small thing we achieved. The first achievement I remember was buying a new stove, which brought us so much joy, and many other achievements followed. Of course, life was not as easy as it may seem. As we navigated our journey, we experienced many ups and downs, joy and sorrow, achievements and setbacks, and a roller coaster of love, tension, happiness, sadness, settling in, and homesickness. Looking back, what makes all these emotions and experiences, positive or negative, special is that we went through them and stayed together. What a blessing! For this, the credit goes entirely to Cherouk, my love, whose kind heart and care for all of us keep our family together. To Cherouk, thank you for everything: for believing in me, for joining me in my ambitions and dreams, for caring for all of us, for everything you do every day to help our family grow, for continuing to believe when times are tough, and for your beautiful heart. I love you.

The blessing that added so much joy to our journey is our sons, the gifts of all gifts, Abdelaziz and Haroun. Abdelaziz was our first experience of everything: becoming a father and mother for the first time, with all the struggles that entails. He was born



one month before we began our journey abroad, and he was our joy and relief when times were tough. His beautiful smile and kind heart gave me strength whenever I needed it. In the middle of the PhD journey, we were blessed with our second gift, Haroun, who brought another dimension of joy to our family. His arrival was another journey in which we had to relearn how to be a father and mother in a home with more than one child, with all the politics that entails! His beautiful smile and affection gave us more strength to continue our journey. To Abdelaziz and Haroun, I love you. You are the meaning of life to me. I pray to Allah that I live to see you become much better versions of us. Lastly, to our upcoming son, I love you even before I see you. We cannot wait to welcome you to the family and receive all the extra blessings you will bring us. As we watched our children grow throughout the journey, we realized that it was not only a learning experience for them, but also for us. Raising a child teaches a person how challenging it is to nurture a young soul. It makes us appreciate everything our parents did, even the things we were not happy with, because we come to understand how difficult it is.

I was also blessed to have a third set of parents: Hussien Abdelkader and Dawlat Aswa, close family friends of my father and mother, who gave me enormous support during the early stages of my independent life and have always been there whenever I needed them. Thank you for everything. I love you. I would also like to thank their wonderful family, Mona, Amr, and Sarah, for being there for us and for all the great times we have shared over the years, and my brother from another mother, Mohamed Hussien, for the lifelong moments we have shared from childhood until now and for all the care and love that were indispensable to me throughout the journey.

One of the greatest privileges of my PhD journey in Delft, and before that my MS in the United States, was meeting the brilliant people whom I came to know and work closely with in academia. Nothing is more fulfilling than building relationships with people who share your ambitions and goals, value hard work, and try to help and support one another.

Throughout this journey, I had the pleasure of working under the supervision of Michaël Wiertlewski, my first promotor and the first person to welcome me to Delft. I still remember our first in-person meeting at the university shortly after I arrived in the Netherlands. It was my first time at the university, and I remember how he took the time to explain what the PhD process looked like, which I understood five years later, and how the transformation from Mostafa to Dr. Mostafa would be a rewarding experience. I appreciated how Michaël was driven by fundamental research ques-

tions, something that I found also drove me. His pursuit of meaningful and impactful research questions profoundly shaped the way I approach research. One invaluable piece of advice Michaël gave me early on was to find the research community to which I belonged. I took this advice so seriously that I attended conferences in many directions, including ultrasonics, physics, tribology, and robotics, until I found my home in the medical robotics community and, to a lesser extent, in physics and tribology. Together, it was quite a journey, like any other, full of ups and downs, but one thing that remained consistent was my respect for him. Thank you, Michaël, for all your guidance and invaluable advice throughout the journey, which shaped who I am today.

I was lucky to be co-supervised by Aimée Sakes, my second promotor, who was a cornerstone of this journey. I still remember the first email I sent Aimée to inquire about the PhD position in Delft and how excited I was to apply. By the time I arrived, she was already on maternity leave. When she returned, she was very enthusiastic and supportive of my progress. I appreciated her willingness to give me the space to explore my own ideas and try things out independently. More importantly, the mental support she gave me during times of uncertainty helped me immensely to keep going. I enjoyed our unconventional, long WhatsApp technical discussions and brainstorming sessions, which led to ideas, some of which became part of this dissertation. I still remember my first trip to the Hamlyn Symposium, which we made together. It was an important step in my PhD trajectory, and I am glad that we shared it. I also remember how she encouraged me and provided opportunities for me to engage with the soft robotics community, which was invaluable to my PhD trajectory. Thank you, Aimée, for believing in me and for all your support and guidance throughout the journey, which shaped who I am today.

I was fortunate to be mentored by Paul Breedveld, my third promotor, who played an indispensable role in this journey. I still remember meeting Paul for the first time in the hallway, where he was printing some papers. I already knew Paul through his papers, so I approached him very enthusiastically, in typical Mostafa fashion, and said, "Hi, I know you." This surprised him because he was not yet aware that a new PhD student was joining the group. He immediately invited me to a meeting in his office, during which he gave me some background on the history of the Netherlands and Delft, something I enjoyed and found very special. Over the years, our relationship grew stronger, although he was not deeply involved in my early years. Things took another turn when we both became excited about a new design idea for a capsule,

following an older design of his that had not worked. This excitement became instrumental to my PhD trajectory, as the idea eventually developed into a complete part of my dissertation. It became clear that we shared a similar taste in design, which made our discussions even more enjoyable. Our exciting idea eventually materialized as a design of which we are both proud, and we have many more ideas to pursue. Apart from work, I enjoyed our long WhatsApp discussions about work and life, and I particularly enjoyed hearing the lessons he had learned from his experiences as a professor and head of the faculty's graduate school. Thank you, Paul, for all the mentorship inside and outside academia and for your trust in me, which shaped who I am today.

I also had the pleasure of being advised by Ron van Ostayen, who played a critical role at the beginning of my PhD by supporting my first research paper, which happened to fall within his area of expertise. I still remember first approaching Ron to discuss my ideas about squeeze-film physics and how welcoming and open he was to technical discussions. These discussions sharpened my research direction and helped me complete the first milestone of my PhD. Our collaboration later continued through the co-supervision of students. Thank you, Ron, for the support that was invaluable to my journey.

At the final stage of my dissertation, I was very interested, and hopeful, in having an examining committee composed of leading experts in the field who could meaningfully assess my work and allow me to engage in deeper technical discussions about it, something I greatly enjoy. I was very fortunate to have such a highly esteemed committee of world-renowned experts join me at this special moment. I would like to sincerely thank Prof. dr. Robert J. Webster III (Vanderbilt University, USA), Prof. dr. Arianna Menciassi (SSSA, Italy), Prof. dr. Jenny Dankelman (TU Delft), Prof. dr. ir. Just Herder (TU Delft), and Dr. ir. Tim Horeman-Franse (TU Delft) for serving as my independent committee members. I am deeply grateful.

Alongside my advisers, there was the community: my friends and colleagues with whom I had the pleasure of sharing this journey. I will start with my great friends in the BITE group office at BME. Kirsten Lussenburg was the first colleague I met, by coincidence, at the office. I greatly enjoyed her spirit and how she was always approachable and enthusiastic about work and life. I was very fortunate to share almost my entire PhD journey with Jette Bloemberg, who was always an inspiring and caring colleague and friend, besides being a great researcher. Her regular check-ins whenever I had not been in the office for a while made a real difference to my

well-being during the journey. Vera Kortman was the innovative soft roboticist who always brought creative, soft ideas to the office with joy and enthusiasm. I greatly enjoyed my discussions with Merle Losch about work and life, and I admired her clear and orderly plans for finishing her PhD, something that someone with a chaotic desk like mine did not have. Esther de Kater was the calm voice in the office who cared about the well-being of others. We shared my memorable first trip to the Hamlyn Symposium, which we both enjoyed greatly. Fabian Trauzettel, my first desk neighbor, I greatly enjoyed our technical discussions about 3D printing and design and always appreciated your presence in the office. Karin Thomassen, our Amersfoort ambassador, I admired your spirit in the office and your enthusiasm for pursuing a PhD alongside all your responsibilities in life, while also being a language learner and Arabic speaker! To the later generation of the office, who joined toward the end of my journey, Nynke, Dina, Giacomo, Haymanot, Jean, and Indra, it was great to spend even a short time together, and I cannot wait to see all the great things you accomplish in your journeys. I would also like to thank the many other brilliant colleagues at BME who made this journey enjoyable: Ludovica, Hassan, Aikaterini, Vahid, Ebrahim, Mathias Peirlinck, Selene Pirola, Suzanne, Bob, Nianlei, Maria, Amir Zadpor, Monika, Abdelrahman, Sara, Remi, David, Mohammad, and many more. Please forgive me if I have forgotten your name.

I was also fortunate to have a second family of colleagues and friends at CoR. My friend Zhaochong Cai, I was very lucky to share so much time with you in the lab, conducting experiments and brainstorming new ideas. I always appreciated your commitment and kind spirit. Laurence Willemet, who shared the entire journey with me, your support, especially at the beginning, was important in helping me continue to believe. Alberico Sabbadini, the first postdoc I worked with, I appreciated your support and advice on navigating the hurdles of a PhD, which were crucial to me at the beginning of the journey. Giuseppe Vitroni, I enjoyed our time together and the chemistry between us, which made our interactions full of joy. Lukas Stracovsky, I enjoyed our lab discussions about global politics and engineering. Kate Karmanova, I admired your social spirit and your contribution to social life at Delft. I would also like to thank the many other colleagues at CoR, including Maximiliane Stoltze, Ebrahim Shahabi, Jagan, Yasemin Vardar, Maurice, Andre, and many more. Please forgive me if I have forgotten your name. I would also like to thank Davood Farhadi from PME for his support and career advice throughout the journey.

Outside TU Delft, I was very fortunate to have great friends and mentors who helped me throughout the journey. Moataz Abdulhafez, my dear friend, coached me from

the first day I began applying for scholarships to pursue my MS abroad and continued to advise me and share all the important moments along the way. Looking back, I am amazed by the amount of time Moataz dedicated to giving me feedback whenever I needed it, always detailed and honest, while he was completing his own PhD. I sincerely thank him for everything. Prof. Yasser Anis and Sayed Metwalli supported and believed in me from the very beginning of my graduate studies and never spared any effort to help whenever I needed it. I sincerely thank them for everything. Prof. Aliaa Rehan gave me unwavering support and always believed in and continue to do so. I sincerely thank her for everything. Prof. Haithem Taha gave me enormous support and advice when I needed it most and provided the example I needed to keep going and believing in myself. I sincerely thank him for everything. Loris Fichera, my MS adviser and friend, guided me during my MS and continued to advise me and check in with me after I began my PhD. I also thank Carla for her kindness and for providing a sense of family away from home. Mohamed Osama helped us enormously in setting up our life in the Netherlands when we arrived. I would also like to thank Mohamed Shaltout, Ahmed Elmahdy, Irene Kuling, Femke van Beek, Krishna Kommuri, Mohamed Fahem, Islam Elsayed, Amr Othman, Abdalla Ali, Amr Hegazy, Ahmed Sharaf, and many more. Please forgive me if I have forgotten your name.

Looking back, the people with whom I shared this journey were its greatest gift and gave meaning to every achievement along the way. The treasure of my PhD was indeed the journey, not the destination.

# Curriculum Vitae

**Mostafa A. Atalla**

02-02-1993      Born in Giza, Egypt.

## Academic Experience

- 2024–present      Postdoctoral Researcher  
University of Twente, The Netherlands  
Robotics and Mechatronics (RaM) Group  
*Advisors:* Prof. dr. ir. Gijs Krijnen,
- 2020–2026      PhD Researcher  
Faculty of Mechanical Engineering, TU Delft  
*Thesis:* When Friction Meets Function  
*Promoters:* Dr. ir. M. Wiertlewski, Dr. ir. A. Sakes,  
Prof. dr. ir. P. Breedveld

## Professional Experience

- 2025–present      Founder  
Mijzan Mechanical, The Hague, The Netherlands  
Engineering consultancy and Product Design Services for Mechatronics Solutions.
- 2020      Intern – Mechanical Design for Robotics and Navigation  
Philips Research North America, Cambridge, MA, U.S.A.  
Image Guided Therapy Department (March 2020–September 2020)
- 2019      Research Intern – Mechanical Design  
Philips Research North America, Cambridge, MA, U.S.A.  
Ultrasound Imaging and Intervention Department (May 2019–September 2019)
- 2018–2019      Product Development Engineer (Intern)  
UbiRos, Natick, MA, U.S.A.

- 2018                      Research and Innovation Engineer III  
Advansys - ESC, Cairo, Egypt  
Innovation Department (March 2018–August 2018)
- 2016–2017              Mechanical Engineer I  
Advansys - ESC, Cairo, Egypt  
Mechanical Department (September 2016–February 2017)

## Education

- 2018–2020              MSc in Robotics Engineering  
Worcester Polytechnic Institute, MA, U.S.A.  
Advisor: Dr. Loris Fichera
- 2009–2014              BSc in Mechanical Design and Production Engineering  
Cairo University, Egypt  
Advisors: Prof. dr. Yasser H. Anis,  
Prof. dr. Sayed M Metwalli

## Honors and Awards

- 2025                      Best Paper and Oral Presentation Award – the 14th Conference on New Technologies for Computer/Robot Assisted Surgery (CRAS)
- 2024                      Maurice-Godet Award for Best Oral Presentation and Paper in Tribology – the 49th Leeds-Lyon Symposium on Tribology
- 2024                      Best Poster Award – BioMechanical Engineering Department Conference (BMeChE) 2024
- 2024                      Best Poster Award – Dutch Soft Robotics Symposium (DSRS) 2024
- 2024                      RoboSoft24 Travel Grant for Excellence – IEEE Robotics and Automation Society (RAS)

# List of Publications

## Publications Related to the PhD Dissertation

### Journal Publications and Manuscripts

5. **M. A. Atalla**, A. S. Sekar, R. van Starkenburg, D. J. Jager, A. Sakes, M. Wiertlewski, and P. Breedveld, "OSCAR: An Ovipositor-Inspired Self-Propelling Capsule Robot for Colonoscopy," (under review). Preprint: doi: 10.48550/arXiv.2602.15309
4. **M. A. Atalla**, D. van Bommel, M. Wiertlewski, A. Sakes and P. Breedveld, "Traversing the Colon: A Review of Self-propelling Mechanisms for Robotic Colonoscopy," (under review).
3. **M. A. Atalla**, J. Nieuwenhuis, A. Martin, X. Wang, A. Candén, M. J. Carré, R. Lewis, A. Sakes, and M. Wiertlewski, "Active Lubrication of Transluminal Medical Instruments," (under review). Preprint: doi: 10.48550/arXiv.2506.07225
2. **M. A. Atalla**, J. J. Tuijp, M. Wiertlewski, and A. Sakes, "Toward Variable-Friction Catheters Using Ultrasonic Lubrication," *IEEE Transactions on Medical Robotics and Bionics*, vol. 6, no. 4, pp. 1375–1381, 2024. doi: 10.1109/TMRB.2024.3464672
1. **M. A. Atalla**, R. A. J. van Ostayen, A. Sakes, and M. Wiertlewski, "Incompressible squeeze-film levitation," *Applied Physics Letters*, vol. 122, no. 24, p. 241601, 2023. doi: 10.1063/5.0149501

### Conference Publications

4. **M. A. Atalla**, D. van Bommel, J. Cummings, P. Breedveld, M. Wiertlewski, and A. Sakes, "Grip as Needed, Glide on Demand: Ultrasonic Lubrication for Robotic Locomotion," in *Proceedings of the 2026 IEEE International Conference on Robotics and Automation (ICRA)*, Vienna, Austria, 2026, accepted. Preprint: doi: 10.48550/arXiv.2602.15608



3. **M. A. Atalla**, J. Nieuwenhuis, A. Martin, A. Canden, X. Wang, M. J. Carré, R. Lewis, M. Wiertlewski, and A. Sakes, "Ultrasonically-lubricated catheters: miniaturization and ex-vivo tissue validation," in *Hamlyn Symposium on Medical Robotics (HSMR) Proceedings*, 2024. doi: 10.31256/HSMR2024.10
2. **M. A. Atalla**, F. Trauzettel, S. P. van Gelder, P. Breedveld, M. Wiertlewski, and A. Sakes, "Mechanically-Inflatable Bio-Inspired Locomotion for Robotic Pipeline Inspection," in *Proceedings of the 2024 IEEE 7th International Conference on Soft Robotics (RoboSoft)*, San Diego, CA, USA, 2024, pp. 586–592. doi: 10.1109/RoboSoft60065.2024.10521968
1. **M. A. Atalla**, J. J. Tuijp, M. Wiertlewski, and A. Sakes, "Ultrasonically-Lubricated Catheters: A Proof of Concept," in *Hamlyn Symposium on Medical Robotics (HSMR) Proceedings*, London, UK, 2023. doi: 10.31256/HSMR2023.12

## Conference Abstracts

6. **M. A. Atalla**, A. Sekar, R. van Starkenburg, D. Jager, A. Sakes, and P. Breedveld, "OSCAR: An Ovipositor-Inspired Capsule Robot for Colonoscopy – A Proof of Concept," 14th Conference on New Technologies for Computer/Robot Assisted Surgery (CRAS), Lisbon, Portugal, 2025.
5. **M. A. Atalla**, T. de Visser, A. Sakes, M. Wiertlewski, and Y. Vardar, "Electroadhesive Stabilization of Soft Robotic Catheters," IEEE 8th International Conference on Soft Robotics (RoboSoft), 2025.
4. **M. A. Atalla**, J. Nieuwenhuis, A. Martin, A. Canden, X. Wang, M. J. Carré, R. Lewis, M. Wiertlewski, and A. Sakes, "Friction modulation of catheters using ultrasonic waves," 49th Leeds-Lyon Symposium on Tribology, 2024.
3. **M. A. Atalla**, F. Trauzettel, S. P. van Gelder, P. Breedveld, M. Wiertlewski, and A. Sakes, "Mechanically-Inflatable Bio-Inspired Locomotion for Robotic Pipeline Inspection," Dutch Soft Robotics Symposium (DSRS), 2024.
2. **M. A. Atalla**, R. A. J. van Ostayen, A. Sakes, and M. Wiertlewski, "On the Squeeze Film Levitation Phenomenon in Incompressible Liquid Environments," APS March Meeting 2023.
1. **M. A. Atalla**, A. Sakes, and M. Wiertlewski, "Near-Field Acoustic Levitation in Liquid Environment," 2022 IEEE International Ultrasonics Symposium (IUS), Venice, Italy, 2022.

## Patents

1. **M. A. A. Atalla**, J. Nieuwenhuis, M. Wiertlewski, and A. Sakes, *Medical device for intraluminal intervention*, Dutch patent NL2036368B1, granted, published Jun. 4, 2025. NL2036368B1

## Other Publications

2. V. G. Kortman, Y. Verberne, **M. A. Atalla**, and A. Sakes, "Advancements in aspiration catheter tip design for thrombectomy: a comprehensive patent review," *Frontiers in Medical Technology*, vol. 6, article 1388638, 2024. doi: 10.3389/fmedt.2024.1388638
1. A. Sabbadini, **M. A. Atalla**, and M. Wiertlewski, "Simple and Fast Locomotion of Vibrating Asymmetric Soft Robots," *Soft Robotics*, vol. 10, no. 6, pp. 1199–1208, 2023. doi: 10.1089/soro.2022.0209



# PhD Portfolio

## Scientific Presentations

### Oral Presentations

- "Grip as Needed, Glide on Demand: Ultrasonic Lubrication for Robotic Locomotion," IEEE International Conference on Robotics and Automation (ICRA), Vienna, 2026.
- "Designing 'With' Friction: A New Paradigm Toward Programmable Physical Interaction," Embodied Intelligence Conference, 2026.
- "OSCAR: an Ovipositor-Inspired Capsule Robot for Colonoscopy: A Proof of Concept," 14th Conference on New Technologies for Computer and Robot Assisted Surgery (CRAS), Lisbon, Portugal, 2025 (Best Oral Presentation and Paper Award).
- "Active Lubrication of Endoluminal Instruments Using Ultrasonic Waves," Dutch Biomedical Engineering Conference, Egmond aan Zee, 2025.
- "Friction modulation of catheters using ultrasonic waves," 49th Leeds-Lyon Symposium on Tribology, 2024 (Maurice-Godet Award for Best Oral Presentation and Paper in Tribology).
- "Ultrasonically-lubricated catheters: miniaturization and ex-vivo tissue validation," Hamlyn Symposium on Medical Robotics (HSMR), 2024.
- "Active Lubrication of Endoluminal Instruments Using Ultrasonic Waves," Health Tech Open Day, TU Delft, 2024.
- "Ultrasonically-Lubricated Catheters: A Proof-of-Concept," Hamlyn Symposium on Medical Robotics (HSMR), London, UK, 2023.
- "On the Squeeze Film Levitation Phenomenon in Incompressible Liquid Environments," APS March Meeting, 2023.

## Poster Presentations

- "Motion-Encoded Friction Anisotropy for Locomotion in Viscoelastic Biological Lumens," Functionality Through Nonlinearity, KU Leuven, 2026.
- "Unconventional Robotic Locomotion: The Ultrasonic Lubrication Way," The Unconventional Robotics Symposium, TU Delft, 2025.
- "Electroadhesive Stabilization of Soft Robotic Catheters," IEEE International Conference on Soft Robotics (RoboSoft), Lausanne, 2025.
- "Toward Variable-Friction Catheters Using Ultrasonic Lubrication," BioMechanical Engineering Department Conference (BMechE), 2024 (Best Poster Award).
- "Mechanically-Inflatable Bio-Inspired Locomotion for Robotic Pipeline Inspection," Dutch Soft Robotics Symposium (DSRS), 2024 (Best Poster Award).
- "Mechanically-Inflatable Bio-Inspired Locomotion for Robotic Pipeline Inspection," IEEE International Conference on Soft Robotics (RoboSoft), San Diego, CA, USA, 2024.
- "Near-Field Acoustic Levitation in Liquid Environment," IEEE International Ultrasonics Symposium (IUS), 2022.

## Invited Talks

- **M. A. Atalla**, "The Body Thinks Too: Physical Intelligence for Medical Robotics – in the Age of AI," ORSI Innotech Surgical AI & Telesurgery Days 2026, Ghent, Belgium.

## Supervision

As daily supervisor and mentor:

- Joan Castelló Belenguer, MSc thesis (2025), "Active Sealing using Squeeze Film Levitation," co-supervised with R. van Ostayen.
- Jack Cummings, MSc thesis (2024), "Ultrasonically-augmented inchworm locomotion," co-supervised with A. Sakes.
- Tim de Visser, MSc thesis (2024), "Stabilization of endoscopes using electroadhesion," co-supervised with Y. Vardar.

- Daan van Bommel, MSc thesis (2024), "Efficient Locomotion for Sliding Robots," co-supervised with P. Breedveld.
- Jelte Nieuwenhuis, MSc thesis (2023), "Ultrasonic friction modulating rings for endoscopes," co-supervised with A. Sakes.
- Rick Matthijssen, MSc thesis (2023), "Multi-sample cardiac biptome," co-supervised with A. Sakes.
- Anand Sekar, MSc thesis (2023), "Design of Intestine Inspection Device," co-supervised with P. Breedveld.
- Jeroen Tuijp, MSc thesis (2022), "Friction Modulation for Endoscopes: A Proof of Concept," co-supervised with M. Wiertlewski.

## Teaching

- Teaching assistant, *Bio-Inspired Design* (graduate course), instructor: P. Breedveld, Fall 2023.
- Grader, *Fundamentals of Biomedical Engineering*, instructor: A. Sakes, Fall 2023.
- Teaching assistant, *Control in Human-Robot Interaction* (graduate course), instructor: M. Wiertlewski, Spring 2021.

## Workshop Organization

- **M. A. Atalla**, S. Sharma, T. Morimoto, T. Nanayakkara, R. J. Webster III, and A. Menciassi, "Soft  $\neq$  Safe? Rethinking Softness in Medical Robotics," 2026 IEEE International Conference on Robotics and Automation (ICRA), Vienna, Austria, 2026. Role: lead organizer.
- N. N. Dei, A. B. Dawood, K. Althoefer, D. Osman, and **M. A. Atalla**, "Soft Robotic Systems for Diagnostics and Therapy in Minimally Invasive and Endoluminal Interventions," the Hamlyn Symposium on Medical Robotics (HSMR), London, 2026. Role: co-organizer.
- B. Calmé, L. Mocellin, **M. A. Atalla**, Q. Boehler, C. Bergeles, T. Horeman, and A. Menciassi, "Smart by Structure: Designing Surgical Devices for Equity and Sustainability," the Hamlyn Symposium on Medical Robotics (HSMR), London, 2026. Role: co-organizer.

- **M. A. Atalla**, L. Lorenzon, M. Maselli, M. Singh, L. Paterno, and K. Suzumori, “Soft Robotics Mimicking Human Body for Medical Applications,” the 8th IEEE International Conference on Soft Robotics (RoboSoft), Lausanne, 2025. Role: lead co-organizer.
- **M. A. Atalla**, M. Finocchiario, Y. Ge, L. Paterno, J. Kinross, M. Abayazid, and T. Nanayakkara, “From Research to Clinical Practice: Bridging the Gap in Robotic Colonoscopy,” the Hamlyn Symposium on Medical Robotics (HSMR), London, 2024. Role: lead co-organizer.
- **M. A. Atalla**, D. Wu, L. Fichera, and A. Sakes, “Bio-inspired medical robotics: from evolution to revolution in healthcare,” the Hamlyn Symposium on Medical Robotics (HSMR), London, 2024. Role: lead organizer.

## Academic Community Service

- Associate Editor, 2026 11th IEEE RAS/EMBS International Conference on Biomedical Robotics and Biomechanics (BioRob 2026).
- Reviewer for *IEEE Transactions on Robotics (TRO)*, *IEEE Transactions on Medical Robotics and Bionics (TMRB)*, *IEEE Robotics and Automation Letters (RA-L)*, *IEEE International Conference on Robotics and Automation (ICRA)*, *IEEE/ASME Transactions on Mechatronics (TMECH)*, *Applied Physics Letters (APL)*, and *Physics of Fluids (PoF)*.
- Volunteer, soft robotics workshop (“Make your own soft robot”), RoboCup 2024 Science Festival, Eindhoven, The Netherlands, July 17–21, 2024.
- Organizing committee member (media team member), World Haptics Symposium, 2023.
- Session chair, American Physical Society (APS) March Meeting, 2023.
- Organizing committee member, Dutch Soft Robotics Summer School and Symposium, 2022.

## Research Outreach | Media Coverage

- TU Delft news feature, “New catheter technology promises safer and more efficient treatment of blood vessels,” November 2024. [Link](#)
- TU Delft research outreach feature, “Designing better catheters – with both more and less friction,” June 2022. [Link](#)

# PROPOSITIONS

accompanying the dissertation

## When Friction Meets Function

Engineering Medical Instruments with Friction to Glide, Grip and Propel

by

**Mostafa A. Atalla**

1. Vibrating interactions create more or less friction. (this dissertation)
2. Alternating contact by coordinating motion can turn friction into function. (this dissertation)
3. Progress is a balancing act between grip and slip. (this dissertation)
4. No friction, no PhD.
5. Reviewer #2 is the proof that not all friction generates traction.
6. Separating results from discussion is a non-effective form of communication.
7. Innovation lies in the friction between chaos and structure.
8. Dutch Bureaucracy is frictionless.
9. Egyptians embrace spontaneity; the Dutch schedule it.
10. You may experience friction reading these propositions.

These propositions are regarded as opposable and defensible, and have been approved as such by the promotors Dr. ir. M. Wiertlewski, Dr. ir. A. Sakes and Prof. dr. ir. P. Breedveld.



